A STUDY OF THE MOTOR UNIT POTENTIAL FOR APPLICATION TO THE AUTOMATIC ANALYSIS OF CLINICAL EMG SIGNALS

by

David Colin Boyd

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Department of Elect. Eng.

The University of British Columbia
2075 Wesbrook Place
Vancouver, Canada
V6T 1W5

Date 19th July, 76

ABSTRACT

A computer model of the human single motor unit potential has been created for the purpose of developing methods of automated analysis in clinical electromyography. This approach was taken in order to examine the effects of pathological changes on the electromyographic potentials.

A comprehensive review of the previous methods of automatic analysis of clinical EMG signals described in the literature has been presented and discussed, together with the relevant work on the production and detection of electrical activity with intramuscular electrodes.

A methodology has been devised for the collection and preprocessing of the electromyographic signals and an EMG data base established at U.B.C. An interactive graphics routine was developed to display the EMG waveform and allow the extraction of single motor unit potentials for further analysis.

A computer model has been proposed for the generation of single motor unit potentials observed during clinical EMG examinations of the normal biceps brachii muscle. This model was based on physiological findings. In the model the single fiber activity was represented by a dipole current source and the motor unit was constructed from a uniform random array of fibers. Motor unit potentials generated from this array were examined at various points both inside and outside the array and the effects of single fiber axial dispersion were investigated. The simulated motor unit potentials generated by the model have been compared with existing data from multielectrode studies in biceps brachii.

The hypothesis that there is a variation in motor unit potential shape at successive discharges was investigated and the model employed for this purpose. It has been shown that for the normal motor

unit potential, one major contributor to the shape variance is electromyographic jitter. The predictions from the model were compared with human experimental data. These results reveal that the variance may be a useful diagnostic indicator, although further research is warranted.

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to my mother

CHAPTER I

INTRODUCTION

1.1 Overview of Electromyography

Electromyography is the detection and recording of electrical activity from a portion of a contracting muscle. It is a method of studying the state of a muscle from the recorded action potentials. The earliest extensive study of the human electromyogram (EMG) was made by Piper in 1912 [1], although human action potentials had already been observed many years previously. Piper recorded potentials during voluntary contraction using surface electrodes and a galvanometer. In 1929, Adrian and Bronk [2] introduced the concentric needle electrode which can be inserted through the skin into the muscle to obtain a more selective picture of the internal activity.

The motor unit is the functional unit of the neuromuscular system and it consists of the anterior horn cell, its axon and all muscle fibers innervated by that axon [3,4]. This is represented diagramatically in Fig. 1.1. The EMG is of proven value to determine malfunction of the motor unit since neuropathic and myopathic diseases cause physiological changes which may reflect in the electrical activity although not *sine qua non*. This is the basis of electrodiagnosis of neuromuscular diseases. For more fundamental information the reader is referred to references on general electromyography and electrodiagnosis [5,6,7,8]. Definitions of the more important terms in Electromyography used in this thesis have been included in Appendix A.

1.2 Generation and Detection of the Clinical EMG

The bioelectric phenomena observed by the electrode are pro-

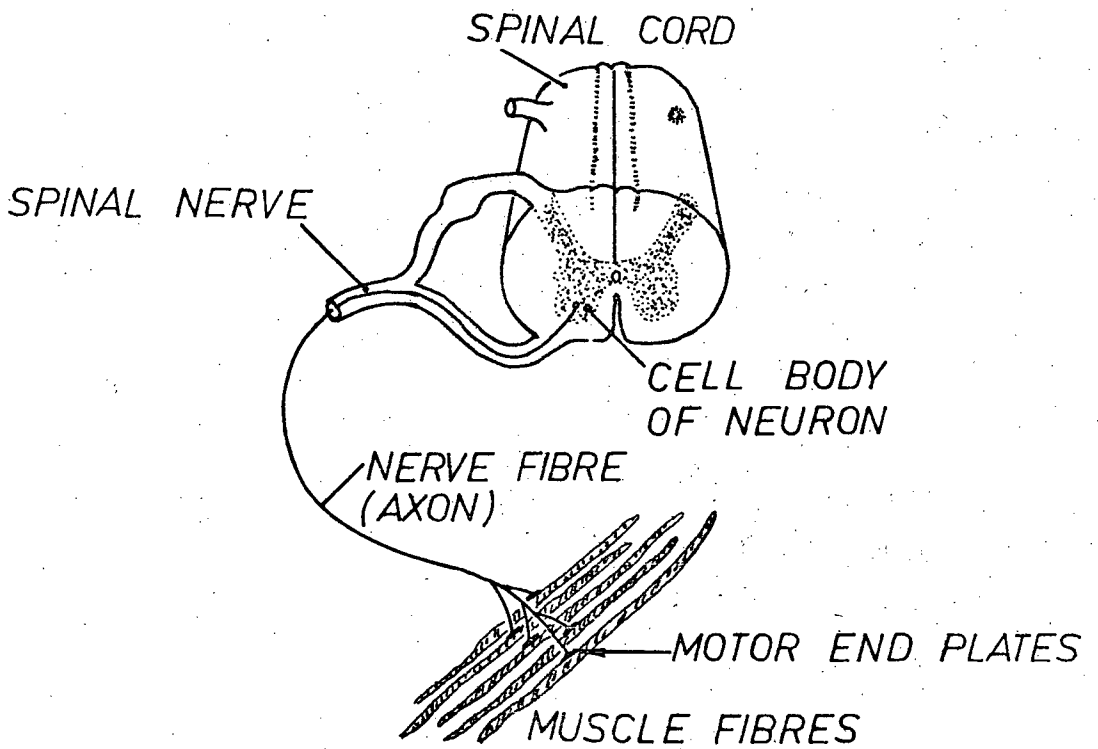

FIG. 1.1 DIAGRAM OF THE SINGLE MOTOR UNIT: THE ANTERIOR HORN CELL, AXON, AND ALL THE MUSCLE FIBERS INNERVATED BY THE AXON.

duced by a summation of electrical events within the muscle. Each active muscle fiber contributes to the production of a time-varying field. The electrode samples this potential field producing an observed action potential. The age, temperature and physiological characteristics of the muscle determine the potential field generated (Buchthal et al. [9]). The distance of an electrode from the fibers composing the motor unit, and the size, shape, configuration and orientation of the electrode are of importance in the time course of the observed signal. Conventionally a concentric needle electrode is used intramuscularly for routine electromyography (Buchthal [10]). Guld et al. [11] have prepared a report on technical factors involved in EMG instrumentation.

According to its geometry and location, the electrode acts as a filter on the motor unit action potential to produce the observed signal (Lindström [12]). With a knowledge of the filter transfer function, signal processing techniques can be employed to select physiological characteristics reflecting the state of the muscle. For example the propagation velocity of the signal along the fibers can be determined by observation of the Fourier power spectra of the electrode signal (Lindström [12]). Literature exists in some of these areas and can be productively employed in design considerations.

Lorenté de Nó [13] determined the potential changes surrounding a single nerve fiber in an infinite, homogeneous volume conductor. Krakau [14,15] extended the analysis by applying Fourier transform techniques indicating the applicability to the radial decline of the potential field in the muscle experiments of Håkansson [16]. In his 1969 monograph, Rosenfalck [17] compared the various models for determining the external action potentials of nerve and muscle in volume conductors, including the

work of Clark and Plonsey [18,19], and extended the analysis to include the effects of muscle fiber thickness and anisotropic media. R.E. George [20] investigated the summation of muscle action potentials for fibers with a Gaussian distribution of axial scatter. George showed that this gives rise to the clinically observed extended duration of a motor unit action potential over a single fiber potential. Based on the expression for the external single fiber action potential derived by Rosenfalck [17], Ekstedt and Stålberg [21] have described the effects of size, shape and orientation of the leading-off surface of a concentric electrode on the observed single fiber action potential.

Using frequency domain analysis, Lindström [12] has determined the relationship between the fiber surface power spectrum observed through a concentric and a surface electrode, and explained the occurrence of dips in the EMG frequency spectrum and the filtering effect of the electrode geometry. Broman and Lindström [22] have described the composition of the single fiber signals into a motor unit power spectrum taking into account the temporal and spatial dispersion of the single fiber action potentials, as well as velocity dispersion among fibers of a motor unit. A model for the summation of motor units in the generation of a power spectrum for intramuscular and surface electrodes has been proposed by Lindström and Broman [22,23].

The application of automatic signal analysis techniques to diagnosis is becoming more common in medicine and is particularly convenient where the basic signal sources are electrical in nature. Such techniques have been applied to the electrocardiogram and electroencephalogram. Much of the present day diagnosis of neuromuscular diseases via the electromyogram is based on a purely subjective examination by the

clinician. Often it is difficult for the clinicians to distinguish between different types of diseases and sometimes between the diseased and normal states. A knowledge of the disordered physiology of the disease state coupled with the transformation induced on the motor unit potentials by the factors mentioned in the preceeding paragraphs, can assist in the selection of automatic techniques of analysis to be applied to the electromyogram.

1.3 A Review of Previous Methods of Automatic Analysis

1.3.1 Special Purpose Analysers

Since the summated action potential recorded from a motor unit includes a potential contribution from each individual muscle fiber, then any fall out of these fibers such as in the myopathies, may reduce the amplitude and duration of the motor action potentials. Even at levels of voluntary activity at which it is not possible to identify individual action potentials, the complex waveform or interference pattern may be affected. Willison [24] in 1963 described a manual method of analysing an EMG record by means of a cursor and a mechanical counting device. This apparatus was used to examine the recordings on 35 mm film which were enlarged by projection. The method involved counting each potential change greater than 100 μV, and recording separately the amplitude of each potential. He was able to demonstrate by this technique those for a patient with muscular dystrophy, the potential changes per second and the mean amplitude of the EMG activity had greater values than that shown by a healthy subject. However, the method was tedious and an automatic electronic analyser was developed by Fitch and Willison [25], to replace the mechanical counter. This analyser was electronic and could be used in real time with the patient. Fitch [26] describes the electronic

circuit in greater detail.

Briefly, in Fitch's circuit Fig. 1.2, the EMG signal from an intramuscular concentric needle electrode is amplified. The output from the amplifer is applied via an RC network to a pair of Schmitt trigger circuits which will respond when a 100 μV positive or negative level appears at the recording electrode. The trigger outputs are fed to a bistable such that it will change state when a positive trigger follows

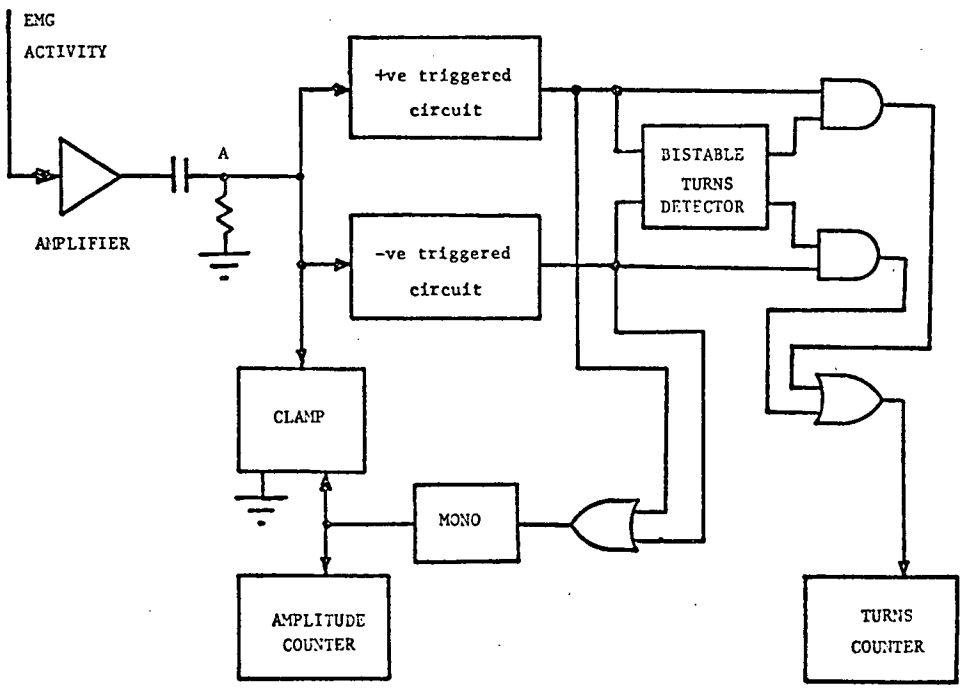

FIG. 1.2 BLOCK DIAGRAM OF ANALYSER USED BY FITCH [26].

a negative one, or a negative trigger follows a positive one. Two AND gates are driven from the outputs of the Schmitt trigger circuits and the bistable. Hence the TURNS counter will only receive a pulse when there is a change in polarity of two successive 100 μV increments of the signal. The operation of either trigger circuit also clamps the point A to zero

volts, thus allowing the next increment of the input signal to be examined. Each operation of the clamp is counted, the total representing the sum of the 100 μV increments.

Rose and Willison [27] used this electronic analyser to quantify the EMG in terms of frequency and amplitude of potential changes during voluntary effort. Standard loads of 2 kg for the biceps, triceps and tibialis anterior and 5 kg for the vastus medialis, were used since the frequency and mean amplitude increased with voluntary effort (Willison [28], Hayward and Willison [29]). The EMG activity of 45 control subjects and 20 patients with muscular dystrophy or polymyositis was analysed. An average of 16 samples for each muscle was taken using a concentric needle electrode. The electronic analyser measured the number of potential changes and the amplitude of every potential change greater than 100 μV (referred to as TURNS COUNT and AMPLITUDE COUNT respectively) per second while the recording was being made. Rose and Willison presented evidence that the frequency of potential changes was often increased in their patients. They also showed that in the cases they examined, the disease was not sufficiently advanced enough that low amplitudes became more important than high counts. Since high count rates might also be shown by patients with chronic partial denervation, it was suggested that the amplitude measurement may help to distinguish between such patients and those with myopathy. When a motor neuron dies collateral branches are sent out from nearby healthy axons. These additional fibers will increase the amplitude as well as the duration of the summated potential (Ermino, Buchthal and Rosenfalck [30]). Thus it seems important to have some measure of the amplitude of a potential.

Later, Hayward and Willison [29] showed that the AMPLITUDE COUNT

measurement could distinguish between chronic partial denervation and myopathy. It was found that even in milder cases of partial denervation, there was an increase in the mean amplitude of the EMG. They also found that the mean amplitude increased as the muscle became weaker, but that such a measurement could not differentiate anterior horn cell disease from peripheral neuropathy.

Dowling, Fitch and Willison [31] used a digital computer together with the analyser to make further measurements on the EMG record (Fig. 1.3).

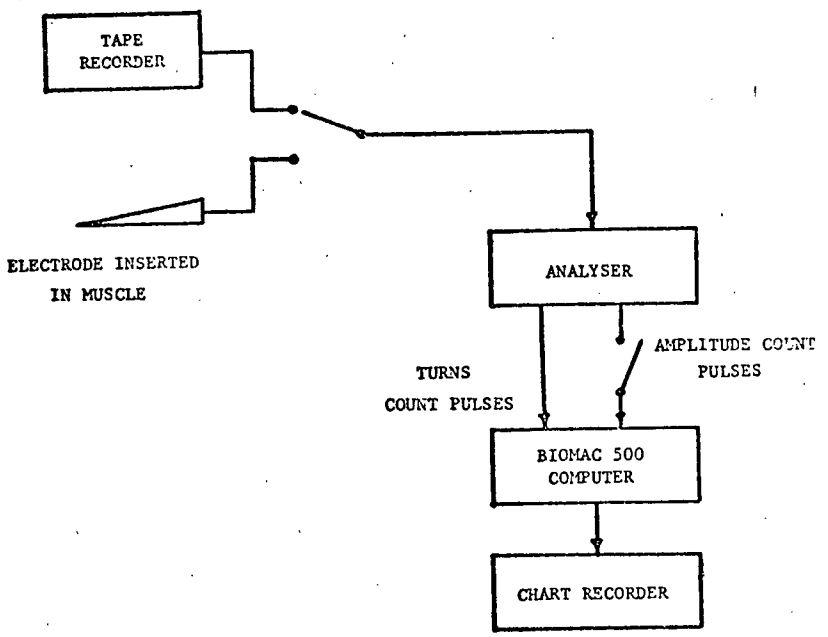

FIG. 1.3 APPARATUS OF DOWLING, FITCH AND WILLISON (1968) [31]

The object of the computer was to accumulate histograms of the time intervals between successive TURNS pulses. The computer used was the Biomac 500 (Data Laboratories Ltd.) as described by Edwards and Aspinall [32]. A TURNS pulse started the sequential addressing of an area in memory at the rate of 64 addresses per millisecond. The next TURNS pulse incremented the content of the memory address reached at that time by one,

and restarted the address sequence. Since addressing intervals are 15.6 μsec, a histogram of the interval between TURNS in sections of 15.6 μsec was built up. Using such a histogram, Dowling et al. revealed that in a patient with dystrophy, there was a marked shift to lower values of TURNS count intervals compared to healthy subjects. Another form of histogram was available as indicated by the switch in Fig. 1.3. The TURNS pulse started the addressing sequence of the memory, while the AMPLITUDE pulses were used to increment and restart the addressing. This histogram provided a means of differentiating between healthy persons and patients with chronic partial denervation.

The work of Hirose and Sobue [33] was similar in principle to that of Willison [24]. Instead of converting the EMG waveform into pulses representing TURNS and AMPLITUDE counts, an analog/digital converter and a JEC-6 spectrum computer (Nippon-denshi) were used to store the actual sampled points as an initial step. The points of each positive and negative peak in the sampled waveform were then selected. Then, the difference between the first negative and positive peak or vice versa and each subsequent pair was calculated. If this difference was less than 100 μV the pair of peaks was ignored i.e. potential changes of less than 100 μV were ignored. A modified waveform was constructed from the significant pairs of peaks and then the number of potentials, the mean amplitude and interval were calculated. All recordings of potentials were carried out under maximum muscle effort. Hirose et al. later compared manual values with this computer analysis (Hirose, Uono and Sobue [34]) and report an application to patients with progressive muscular dystrophy (Hirose, Uono and Sobue [35]).

The duration of the individual motor unit potential is often a

good indicator of disease states, (Buchthal et al. [9]) but the individual motor unit potentials are often difficult to distinguish at high levels of contraction. Moosa and Brown [36] used a purely analog analyser to detect indirectly changes in the action potential duration with maximum strength of contraction in the muscle. This analyser was used to obtain an index representing the reciprocal of the mean phase duration. The index was based on a similar one, ϕ, presented by Van den Bosch [37], where

$$\phi = \frac{S_1+S_2\ldots S_n}{T_1+T_2\ldots T_n} = \frac{1}{\text{Mean duration of the phases}} \tag{1.1}$$

and T_n, S_n are the duration and the number of positive and negative peaks respectively in the nth deflection of the waveform. Moosa and Brown avoided the problem of distinguishing between action potentials and small noise deflections by weighting the mean duration according to the amplitudes of the deflections. Hence,

$$\psi = \frac{\sum_{n=1}^{n=N} a_n}{\sum_{n=1}^{n=N} a_n T_n} \tag{1.2}$$

where ψ is the symbol used for the index, and a_n is the amplitude of the nth deflection of the waveform. If it is assumed the waveform p(t) is sinusoidal, they show ψ can be represented by

$$\psi = \frac{\int_0^T |\dot{p}(t)|\, dt}{\pi \int_0^T |p(t)|\, dt} \tag{1.3}$$

A simple analog computer circuit implementing this expression was the basis of the analyser and the value for ψ could be continuously displayed

by a chart recorder. Moosa and Brown claimed that this index was nearly independent of the strength of contraction of the muscle. The index could differentiate between normal and myopathic EMG signals.

Moosa, Brown and Dubowitz [38] applied the analyser to carrier detection in Duchene type muscular dystrophy. The index was significantly higher in a proportion of the cases examined. However, they advise the estimation of serum creatine phosphokinase to supplement this EMG analysis to give an 80% rate of carrier detection.

In contrast to the previously discussed methods, Lang and Vaahtoranta [39] used the principle of sampling and averaging voluntary EMG activity to create an average motor unit action potential. The electrode was kept at a constant site in the muscle and the sum of different motor unit potentials (MUP'S) obtained. This sum they termed the averaged muscular potential (AMP). The method is illustrated in Fig. 1.4.

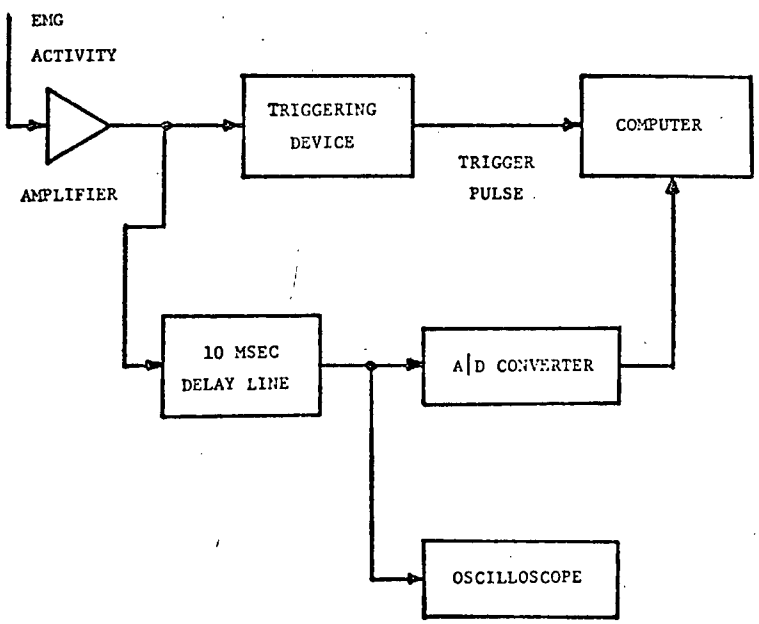

FIG. 1.4 METHOD OF LANG ET AL. TO OBTAIN AN AVERAGED MUSCULAR POTENTIAL (A.M.P.)

The EMG signal was amplified and connected to a device which would trigger the computer when the rapid negative phase satisfied the amplitude condition (see Lang, Nurkkanen and Vaahtoranta [40]). The signal was then fed through a 10 msec hardware delay, such as that described by Nissen-Petersen et al. [41], to the oscilloscope and to the analog/digital converter of the computer. Thus the 10 msec of the signal before the trigger was recorded. Single MUP's were recorded during weak muscle contraction. Normally about 100-200 signals were summed with the computer set at 200 μsec address time. Advantages of this method are that the parameters such as shape, duration and amplitude are averaged, and that further computer analysis could be employed since the signal is in a digital form.

Lang and Tuomola [42] showed very clearly that the AMP recorded from neuropathic, myopathic and normal muscles significantly differ. The signal/noise ratio is improved and the baseline preceding the signal becomes more exact. Parameters dependent on the reliability of a baseline may then be more easily defined.

Another special purpose EMG analyser was described by Kunze [43]. Briefly, the analyser transformed the action potential into pulses for the different measurements: potential duration, polarity of the first phase, number of phases and the amplitude integral. The pulses were then counted and x-y plots could be made for the following parameters while isometric contraction was varied: the amplitude integral versus the number of phases; potential time (the sum of the pulses in all valid potential durations within the analysis interval) versus the number of phases; potential time versus amplitude. Using such plots Kunze then showed the difference in patterns of a patient with myogenic muscle disease (poly-

myositis) and one with a neurogenic muscle disease (neural muscular atrophy), and how these patterns differed from those in the normal control group.

Frequency analysis of the EMG waveform may also provide information about the state of a muscle. It has been shown by Larsson [44] using a frequency analyser of the type described by Kaiser and Petersén [45,46], and by integrating the output from the filters, that in myopathies there was an increase of activity in the high frequency filters as compared to normals. Larsson also pointed out that denervated muscles tended to show a decrease in the output at high frequencies. Larsson [47] presents further evidence of this in a later paper and indicates that the frequency spectrum depends on the duration of the symptoms.

All of the above methods use electronic circuits to detect special features of the EMG waveform. Some investigators, however, use the computer directly as a tool to measure EMG characteristics as will be discussed in the next section.

1.3.2 Measurement Analysis by Computer

Many of the characteristics of muscle action potential described by Buchthal, Guld and Rosenfalck [48], are affected by disease (Kugelberg [49], Pinelli and Buchthal [50], Buchthal and Pinelli [51,52]). The actual measurement of these parameters proved to be laborious and time consuming. Fortunately they are suitable for computer automation. A common difficulty, however, is the automatic detection of the onset and end of the action potential. This problem was approached in several different ways by various investigators.

Rathjen, Simons and Peterson [53] used a minicomputer (Digital Equipment Corp. PDP-8) to measure and print out the duration parameter.

Single motor unit potentials were picked up by a bipolar coaxial electrode and amplified. An analog/digital converter then sampled the signal and a computer program examined each digitized sample to look for valid durations. A valid duration existed if the signal had exceeded an amplitude threshold or trap level and returned to the trap for a certain number of samples.

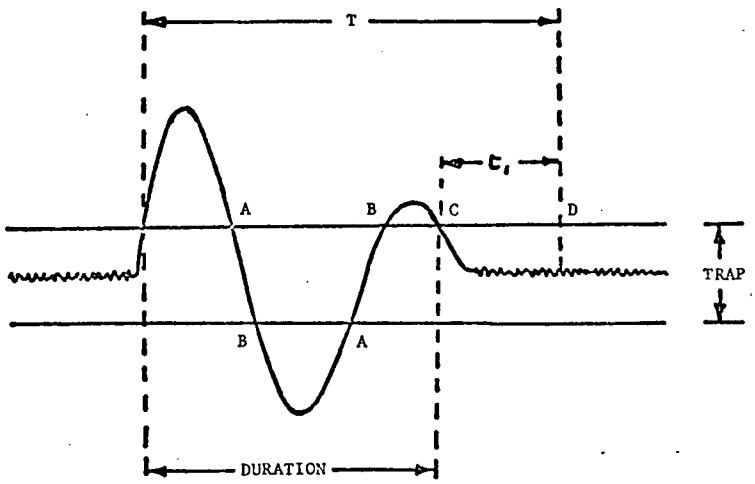

FIG. 1.5 METHOD USED BY RATHJEN ET AL. (1968) TO MEASURE DURATION

In any time PERIOD AB there are not enough samples within the trap to terminate sampling. This only happens during the time PERIOD CD. Thus the duration is given by $T-t_1$.

This is indicated in Fig. 1.5. The purpose of this trap was to eliminate baseline drift and low amplitude noise. Once the signal had left the trap, an internal clock in the computer was started, and stopped only when the program had found a valid return. Further checks were made on the duration. The program established if it was greater than a minimum duration. This was to eliminate high frequency noise potentials. Also, the amplitude had to be larger than a preset value to avoid distant motor unit potentials. There was no discussion of the accuracy of the method reported.

Automatic recording of histograms of the motor unit potential duration was developed by Hausmanowa-Petrusewicz et al. [54] and Kopeć

and Hausmanowa-Petrusewicz [55,56] to study as large a number as possible of the potentials. Kopeć and Hausmanowa-Petrusewicz [55] applied the Polish Averaging Computer, ANOPS (Analyser of Perodic Noise discharges) for this purpose. The amplified EMG was fed to special input circuits before entering the computer, as shown in Fig. 1.6. These input circuits

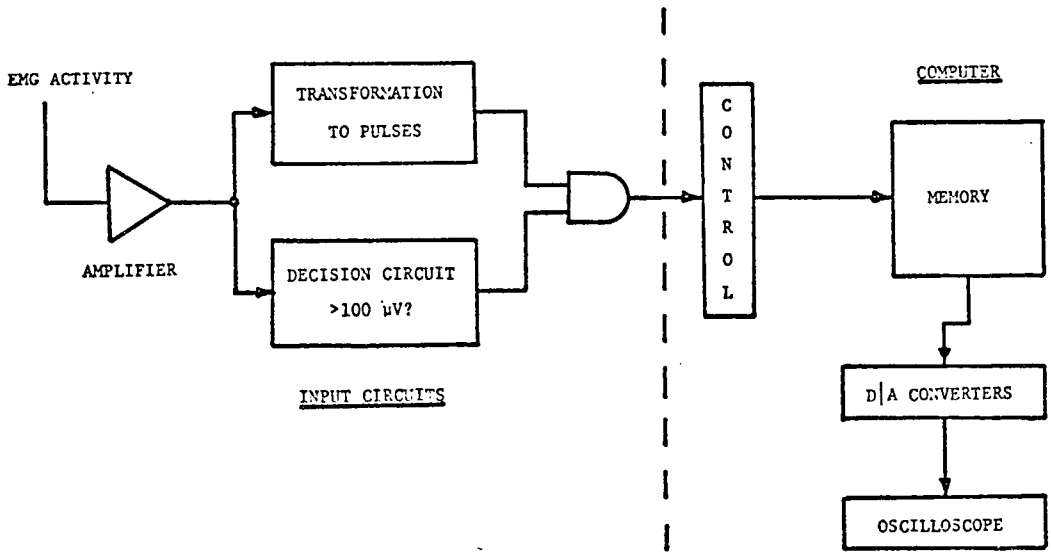

FIG. 1.6 BLOCK DIAGRAM OF THE APPARATUS AS DESCRIBED BY KOPEĆ AND HANSMANOWA-PETRUSEWICZ [55]

transformed the motor unit potentials into rectangular pulses equal in width to their duration at a 20 μV level. A decision circuit 'AND' gated only those potentials with absolute amplitudes higher than 100 μV, which were then fed in pulse form to the computer. In the computer the pulses were summed and stored in memory locations according to their duration. When the histograms were to be displayed, the content of memory was output to a digital/analog converter which was connected to the vertical and horizontal amplifiers of the computer oscilloscope. The vertical bar represented the number of motor units of that memory location and the horizontal corresponded to the duration of the measured

pulse in 0.8 msec intervals. A pilot study was carried out on a number of subjects including healthy ones, various cases of myopathy and cases of neurogenic muscle atrophy. Histograms of the healthy subjects showed peaks in the range 6 to 12 msec with the spectral width, estimated at the 50% level of the maximum height, of 4 to 15 msec. In the cases of myopathy there was a definite shift to the left of the histogram with peaks lying from 3 to 6 msec and the spectral width from 2 to 11 msec. On the other hand, in cases of neurogenic atrophy the histograms were shifted to the right having peaks in the range 8 to 19 msec and a spectral width of 5 to 26 msec.

The above method was later extended by Kopeć, Hausmanowa-Petrusewicz, Mawski and Wolynski [57], to include histograms of the number of phases per unit time.

Lee and White [58] used the slope to determine the beginning and end of a computer-identified motor unit potential for measuring duration and peak to peak amplitude. A number of sampled points was examined by a PDP-12 computer for a voltage change exceeding a preset level. When the slope exceeded this limit, the beginning of that section was marked as the onset of the potential. As most potentials have a gradually sloping tail, a different slope was used to find the end. A display of the potential, together with measured parameters and cursors indicating the onset and the end, was also available which allowed the operator to check the computer. Lee and White's clinical result on one patient with polymyositis supported the work of Kopeć, although the method of duration computation differed.

Computer analysis described previously have certain limitations. There is no computer check that identified potentials are of one single

motor unit, and thus artifacts and potentials due to a superposition of two different motor unit potentials, are not always rejected. Also, the part of the signal that occurs before the triggering level is reached will be lost.

Bergmans [59,60] used a software delay line which would allow values of the signal to be kept ahead of the trigger. To be sure the potentials were representative of only one single motor unit, a computer program required that the waveform should occur twice before its parameters were measured. The parameters of duration, amplitude, polarity of the initial phase, number of phases and the number of peaks were then measured. The EMG activity was recorded during slight voluntary contraction, amplified and sent to an analog/digital converter connected on-line to a PDP-12 computer. The first function of the computer was to isolate one potential out of the EMG record using an interactive display to adjust a threshold criterion. The second function was to identify, if possible, subsequent potentials as belonging to the same motor unit and average the stored and new potentials together to improve the signal-to-noise ratio. The third function was to recognize that an isolated potential did not belong to a stored average motor unit potential and to generate a new motor unit pattern in storage for identification with subsequent isolated potentials. The final function was that of computation and display. Usually 5 different motor unit potentials were obtained and each displayed in turn after parameter measurements were made. In this method, duration was measured between the first n consecutive points exceeding a threshold and the last m consecutive points exceeding the same threshold. The interactive display again allowed the operator to adjust these points or to verify duration had been measured correctly. Histograms of the parameters could later be generated.

In the EMG analysis described by Bergmans, there was close interaction of not only the operator with the computer but also between the patient and the computer. In another program developed by him, the task of recognizing the different motor unit potentials was left solely to the operator who rejected or accepted a potential. Thus, further intervention of the operator meant that the errors were reduced.

1.4 Scope of the Thesis

1.4.1 Thesis Objective

Few of the methods of automatic analysis of the electromyogram described above, with the exception of Willison's, are clinically practical. Motivation does exist to develop clinically practical and reliable automatic methods as an adjunct to normal clinical assessment of neuromuscular diseases and integrate these with existing equipment. Boyd, Bratty and Lawrence [61] have proposed a number of design features required by such a system. Before a system is realized however, a basic understanding of the spread of electrical activity in the tissues, electrode properties and the pathophysiology of the diseased state, is required. None of the automatic methods discussed in section 1.3 were based on a quantitative relationship to the physiological effects of disease on electrical activity. The approach that was taken here was to construct a model of the single motor unit as seen by an electrode in the muscle. This model was then compared to physiological data that was recorded from normal motor units in the human and also to the data of other investigators. By using this model changes in the diseased state could be predicted. To the knowledge of the author, this approach has not been taken previously. Thus the overall objective of the research described in this dissertation is to contribute to our understanding of the human

motor unit potential in the normal and abnormal states in order to shape the development of a clinical system.

1.4.2 Thesis Outline

A data base has been established containing normal and pathological EMG activity. The method used to collect this data is the subject of Chapter II. This chapter also describes the use of interactive graphics to isolate single motor unit potentials from the recorded EMG activity for averaging and further processing.

Chapter III describes the development of a computer simulation model of the human motor unit in the biceps brachii muscle. This model is an attempt to understand the single motor unit potential and how the electrode affects this potential. The model is compared with the real data and with other clinical investigations given in the literature.

It has been observed clinically that the shape of the motor unit potential varied in subsequent firings in the diseased state [62]. Chapter IV investigates this hypothesis by studying the variance in shape of the single motor unit potential at each firing. The model developed in Chapter III is used to show the effect of 'jitter' of the single fiber potential on the variance in shape of the motor unit potential.

A summary of the contributions of this research together with conclusions is given in Chapter V. Suggestions for further work are also outlined in this chapter.

A number of appendices are included for reference purposes. Appendix A contains definitions of some of the more important terms used in electromyography. Appendix B is a copy of the 'EMG RECORDING - INFORMATION SHEET' which was used during the analog recording of the EMG data. Documentation for the interactive graphics program has been included in Appendix C for the use of other researchers, and Appendices D and E con-

tain the computer program listings for the simulation and plotting of single motor unit potentials.

CHAPTER II

THE ACQUISITION AND PREPROCESSING OF THE EMG DATA

2.1 Introduction

In this chapter the acquisition of the EMG and the subsequent preprocessing used to test the models developed in later chapters is described. Samples of EMG activity, at low level contraction, from both normal and pathological muscle were recorded in analog form and digitized for further analysis on the computer. The objective was to obtain action potentials from a single motor unit.

Action potentials were recorded from the biceps brachii muscle which is the principle flexor of the elbow joint [63]. This muscle was chosen as it was easily identified and examined. In the normal electromyogram recorded from this muscle, the single motor unit potential can be bi- or triphasic, with monophasic potentials being less common [64]. Polyphasic potentials which contain more than four phases occur in approximately 4% of the motor units. For a weak volitional effort, the frequency of firing of a motor unit potential is between 5-15/sec. With increased effort, the rate of discharge increases and also additional motor units are recruited. The observed amplitude of the action potentials recorded intramuscularly with a concentric electrode ranges from a few microvolts to about 5 mV, with a total duration of 3-16 msec in the normal biceps brachii [9]. The duration of the positive to negative deflection is in the order of 100-200 μsec. Typical EMG activity recorded from a normal biceps brachii of a male subject at a low muscle contraction level is shown in Fig. 2.1 (a).

Recordings were also made of patients with polymyositis. This is a muscle disorder in which there is muscular weakness and is one of

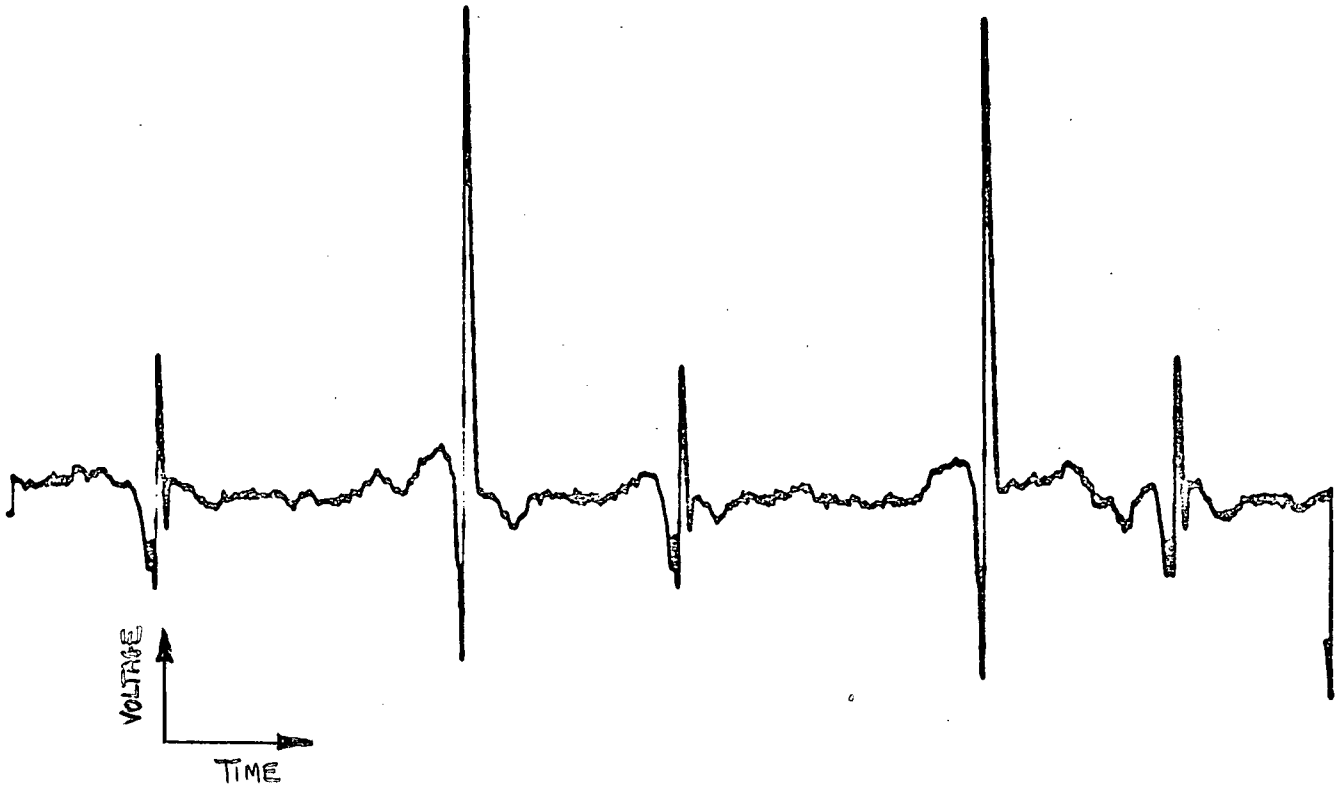

FIG. 2.1 (a) TYPICAL SINGLE MOTOR UNIT POTENTIALS RECORDED FROM THE NORMAL HUMAN MUSCLE (BICEPS BRACHII)

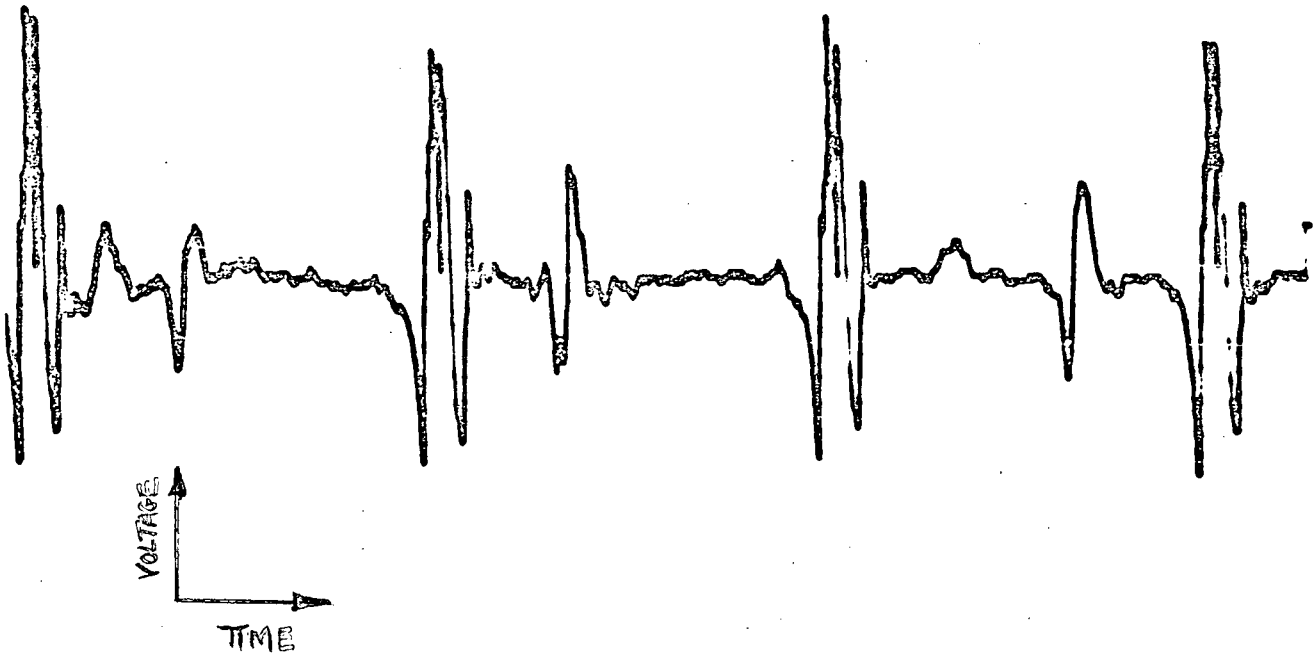

FIG. 2.1 (b) ABNORMAL POTENTIALS RECORDED FROM A PATIENT WITH POLYMYOSITIS (BICEPS BRACHII)

the most frequently occurring primary myopathies in adults. The motor unit potentials are polyphasic and usually of shorter duration than the normal. As electromyographic changes in polymyositis are patchy in distribution within the muscle, the muscle must be explored to uncover abnormal potentials before recording. The EMG activity recorded from such a diseased biceps brachii muscle is shown in Fig. 2.1 (b).

Section 2.2.1 of this chapter describes the method used for analog recording of this EMG data while section 2.2.2 outlines the procedure used for digitization.

Single motor unit potentials may be isolated from the recorded EMG activity due to the following considerations:

(1) At low levels of muscle contractions only a few motor unit potentials fire to maintain the contraction.

(2) By slight movement of the electrode under the condition given in (1), the clinician can move to within the pick up range of one unit.

(3) Often the subject while viewing the oscilloscope trace of his electrical activity can isolate a motor unit potential by varying the contraction level of his muscle.

(4) Motor unit potentials firing within the range of the electrode can usually be distinguished by a characteristic shape.

(5) Since each motor unit fires repetitively and usually out of synchronism with other motor units, a single motor unit potential can be identified.

A method has been developed using interactive graphics to extract identified action potentials from the digitized EMG activity. This method is described in Section 2.4.1.

For further analysis, it is required to form the averaged motor unit potential. Before averaging, the extracted single motor unit potentials must be aligned in the formation of an ensemble. The procedure for alignment of these potentials is discussed in Section 2.4.2.

2.2 Recording and Digitization of the Data

2.2.1 Analog Recording of EMG Data

A data base consisting of the EMG activity recorded from normal and diseased muscle has been established for use in this research. Recordings were made at the Vancouver General Hospital under the supervision of an experienced neurologist[1]. Fig. 2.2 (a) shows the equipment used during the data acquisition. A DISA, 3 channel Electromyograph Type 14A30 was used to amplify and visually display the EMG activity. The filters on the DISA Electromyograph were set with the high pass at 20 Hz, to avoid shifting of the baseline due to electrode movement, and the low pass filter at 1 KHz thereby reducing high frequency noise components. A Hewlett-Packard 3960A FM instrumentation tape recorder was connected to the electromyograph and used to record the EMG activity on 3M TYPE 871 instrumentation tape at a speed of 15 i.p.s. The electrode used was the standard concentric needle electrode used for intramuscular recording (DISA ELEKTRONIK 13K51). The diameter of the needle was 0.65 mm, the inner conductor being 0.07 mm^2. This electrode has been represented diagramatically in Fig. 2.2 (b).

Standardization of the electromyographic examination cannot be complete as in electroencephalographic examinations, for example, because standard electrode positions are not applicable. Though recordings

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1 Dr. P.J.A. Bratty, Clinical Associate Professor of Medicine (Neurology), University of British Columbia.

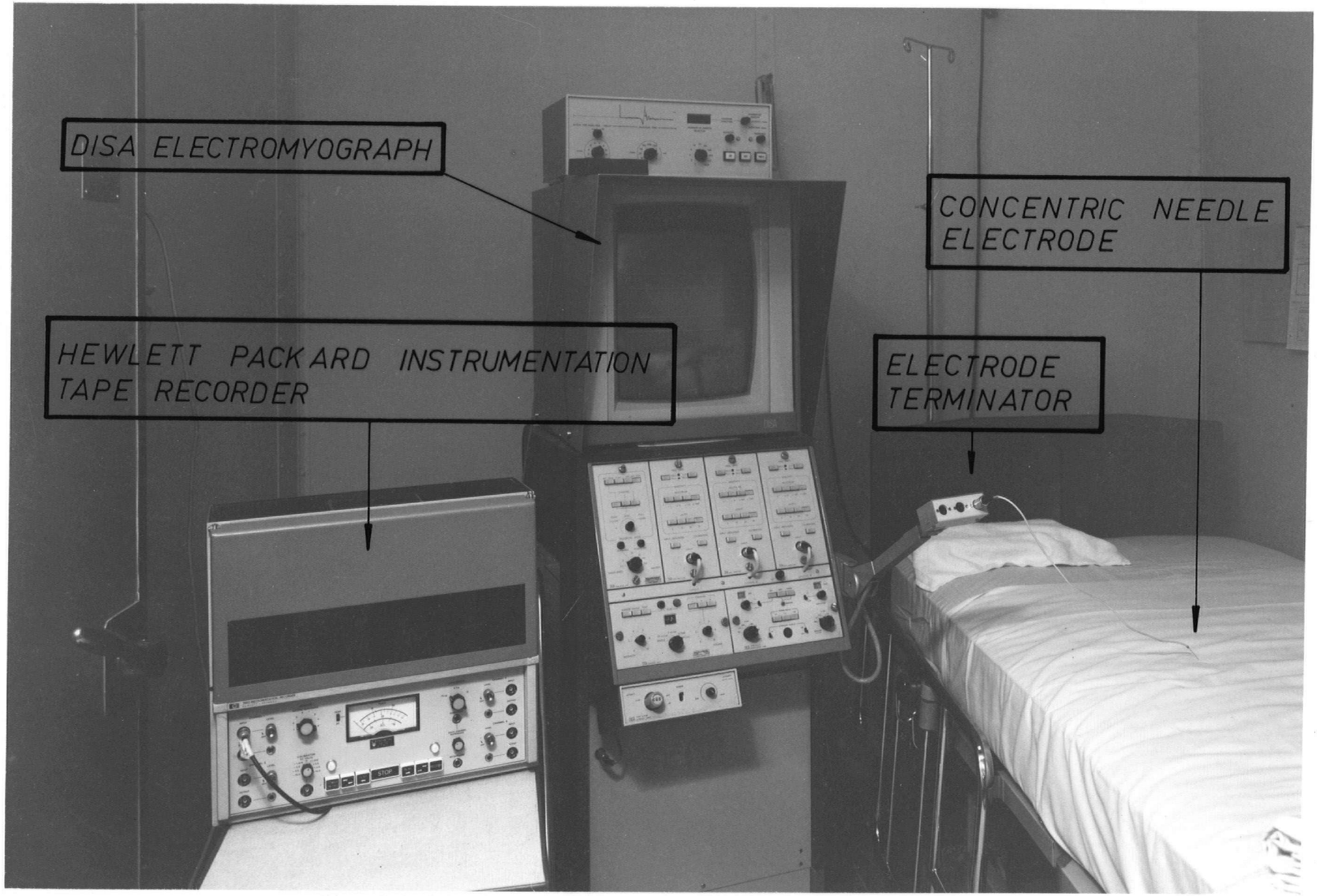


Fig. 2.2 (a) THE EQUIPMENT USED FOR DATA ACQUISITION

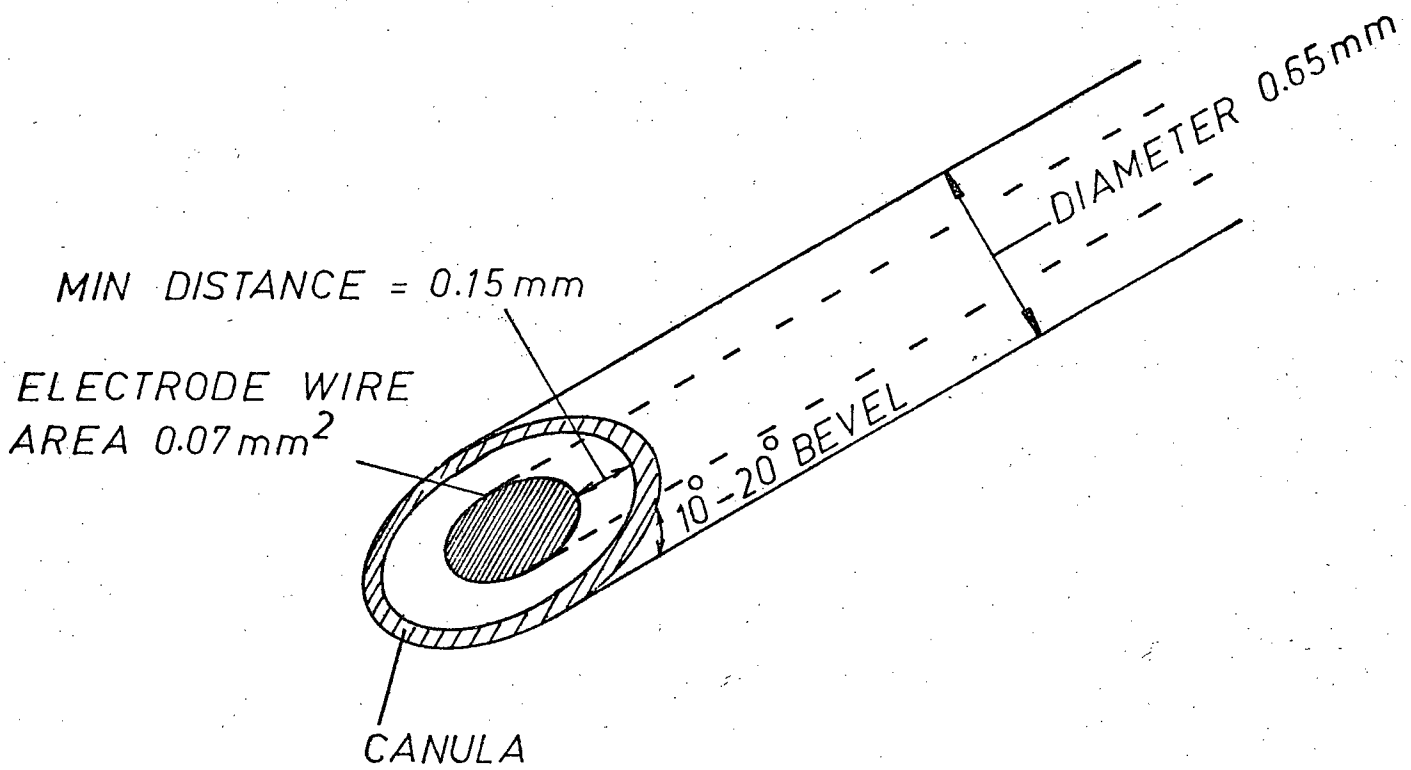


Fig. 2.2 (b) DIAGRAM OF A CONCENTRIC NEEDLE ELECTRODE

This type of electrode was used for recording the EMG.

were usually taken from the biceps brachii muscle, no attempt was made to standardize the position of the electrode in this muscle, or the place of insertion, except that the innervation zone and ends of the muscle were avoided.

The EMG activity was recorded in the following manner. The subject was asked to make the weakest possible muscle contraction and by slowly changing the position of the needle electrode, a point was reached at which it was possible to record a single motor unit potential or at least one of considerably larger amplitude than the others. The gain of the DISA Electromyograph was altered to give a signal 0.5 volts peak to peak. Changes of the electrode position and details of the recording equipment settings were noted in accordance with the 'EMG RECORDING - INFORMATION SHEET', shown in Appendix B.

2.2.2 Digitization of the EMG Waveform

The recorded EMG activity which was stored in analog form was digitized for further processing using the Data General Nova 840 system at the U.B.C. Electrical Engineering Department. This was accomplished by first reproducing the recorded EMG activity on the tape recorder at the lower speed of 15/16 i.p.s., to reduce the effective sampling rate by a factor of 16. The signal was then low pass filtered at 62.5 Hz with a filter gain of 20 dB using the Krohn-Hite 3342R filters and subsequently digitized. The digitized samples were then stored on 9-track IBM compatible magnetic tapes by using the Nova 840 computer system. The analog waveform was sampled at a frequency of 512 Hz (giving a factor of >8 improvement over the Nyquist sampling rate), and converted to a 12 bit binary number. Although the resolution was 12 bits, each sample was stored on the magnetic tape as two 8 bit bytes, right adjusted

with the other 4 bits being a copy of the sign bit (2's complement format). This simplified the programs described in later chapters. An end-of-file mark on the magnetic tape was established to indicate a change of subject, or a change of the electrode position in the same subject. This was in accordance with the 'EMG RECORDING - INFORMATION SHEET' (see Appendix B).

2.3 EMG Data Base

Sample EMG activity was obtained from the biceps brachii muscle of three normal male subjects and two female subjects. Several different motor unit potentials were selected for recording from each subject. Recordings of EMG were also made from the same muscle of one male and one female with previously established diagnoses of polymyositis. Only abnormal EMG was recorded from these two patients.

To test predictions made in Chapter IV, EMG was recorded from a patient with Myasthenia Gravis, a neuromuscular transmission disease.

Examples of some of the recorded activity are shown in Fig. 2.1 (a) and Fig. 2.1 (b).

2.4 The Averaging of Single Motor Unit Potentials

2.4.1 The Interactive Graphics Routine

To extract single motor unit potentials from the time series EMG waveform an interactive graphics routine known as INTERACT was developed for use at the Adage Graphics Terminal, U.B.C. Computing Center. Documentation for the use of this program is given in Appendix C. This method was adopted for several reasons:

(1) Artifacts, such as distant motor unit potentials, the superposition of two motor unit potentials or an unsteady baseline due to electrode movement, could be eliminated by visual assessment.

(2) By storing one motor unit potential on the screen, and by superimposing another potential on it, a decision could be made on the basis of shape if the second potential was from the same motor unit.

(3) The portion of the motor unit potential to be analysed could be chosen visually and other potentials matched to have the same time period.

The EMG waveform was displayed on the screen, one record at a time (4096 bytes or 2048 sample points), being read directly from the digital magnetic tape. A single motor unit potential was then selected, centered on the screen and enlarged. The result of this procedure is shown in Fig. 2.3 (a). The section of the potential to be analysed was visually chosen by means of hairline cursors (Fig. 2.3 (b)), and the potential extracted and stored on magnetic disk for further processing. The selected potential was also 'stored' on the screen and other action potentials from the same motor unit matched to it. An ensemble of single motor unit potentials was formed in this manner.

2.4.2 Averaged Motor Unit Potential (AMUP)

As stated previously, single motor unit potentials were aligned visually using the interactive graphics routine (Appendix C). To reduce alignment error, the mean square difference between the two potentials was minimized. This was achieved by the following method. A window of N-10 sample points, where N is the total number of sample points of the action potential was considered, such that up to 5 sample points on each side of this window could be moved into it, or out of it.

The coefficients were then computed:

$$RR(k) = \frac{1}{N} \sum_{i=1}^{N} (x_i - y_{i+k})^2 \qquad (2.1)$$

k = 0, 1 ... 5 sample points.

and

$$RL(k) = \frac{1}{N} \sum_{i=1}^{N} (x_{i+k} - y_i)^2 \qquad (2.2)$$

k = 0, 1 ... 5 sample points

where N is the number of sample points, $\vec{y}(1, \ldots N)$ is the potential to be aligned, and $\vec{x}(1, \ldots N)$ is the potential to which $\vec{y}$ is aligned.

The

MINIMUM (RL(k), RR(k)) k = 0, 1 ... 5

was chosen and the potential y moved k sample points in the direction of the minimum coefficient. It was found that the visual method of alignment satisfied the mean square difference criterion within 1 or 2 sample points. Finally an averaged motor unit potential was computed from the ensemble of aligned potentials.

Fig. 2.3 (c) summarizes the procedure for obtaining the averaged motor unit potential from the digitized EMG waveform stored on magnetic tape.

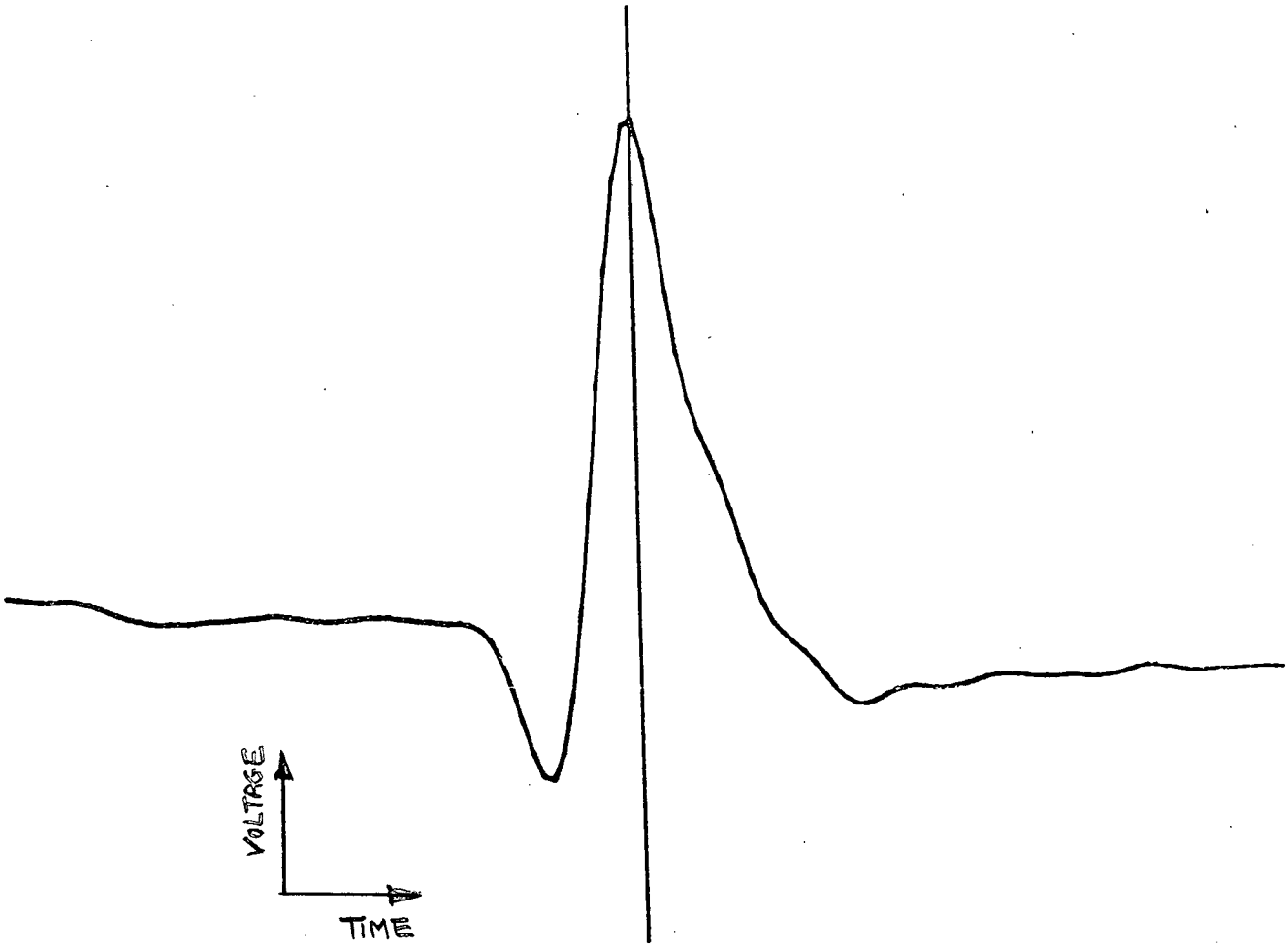


FIG. 2.3 (a) SELECTION, CENTERING AND ENLARGEMENT OF A SINGLE MOTOR UNIT POTENTIAL.

A single motor unit potential has been selected from the EMG waveform and enlarged on the screen. The peak of the potential is centered on the screen by means of the x hairline cursor (the vertical line).

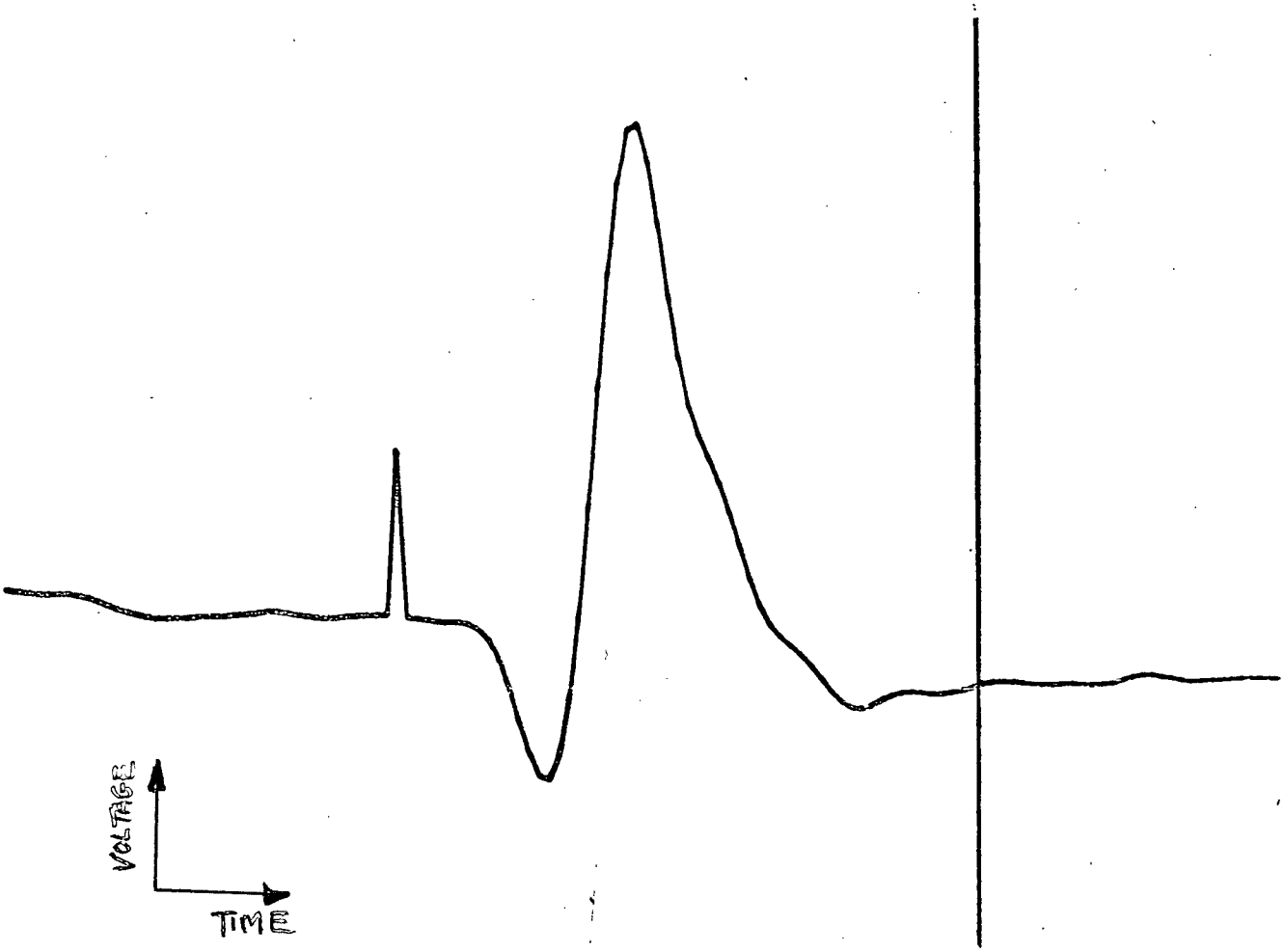


FIG. 2.3 (b) CHOSING THE PORTION OF THE POTENTIAL TO BE EXTRACTED FROM THE WAVEFORM FOR FURTHER PROCESSING

The x hairline cursor (the vertical line) is used to select the beginning and end of the action potential. In this figure, the beginning of the potential has been chosen and marked by a pulse.

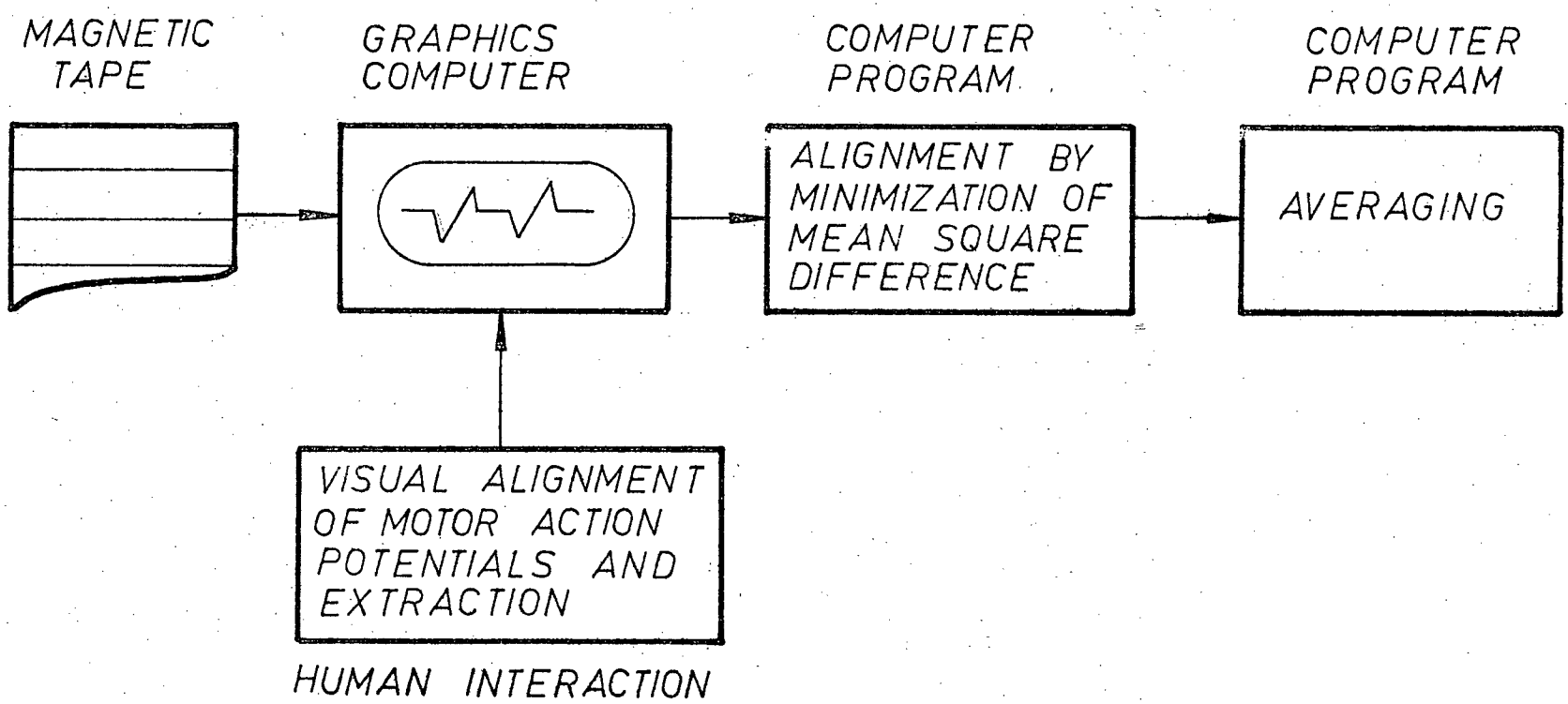


FIG. 2.3 (c) AVERAGING OF A MOTOR UNIT POTENTIAL

CHAPTER III

A MODEL FOR THE GENERATION OF SINGLE MOTOR UNIT POTENTIALS

3.1 Introduction

Electromyography has achieved widespread application as an aid to the diagnosis of neuromuscular diseases. In many of the standard methods of clinical EMG examinations, it is the shape of the action potential that is analysed. For this purpose the study is carried out with the EMG signal recorded at low contraction levels of the muscle, usually such that only one single motor unit potential is distinguishable. In order to give a better understanding of the form of the action potential that is recorded from a motor unit, a simulation model can be developed. Such a model may be applied to study how disease mechanisms affect the recorded motor unit potential.

A motor unit action potential is the result of a summation of electrical activity within the muscle. Each active fiber belonging to the motor unit contributes to the production of a time varying potential field. The time course of the single fiber potential thus determines the overall motor unit potential. Many studies, both experimental and theoretical, of the single fiber action potential of the nerve and muscle have been presented in the literature.

3.1.1 Single Fiber Action Potential Models

Lorente de Nó [13] showed mathematically that the potential at any point in an infinite volume conductor surrounding a nerve fiber can be related to the membrane current density distribution. He showed that the form of the volume conducted action potential was triphasic in the nerve. He also described how the membrane current density can be determined by experimental measurement of the second derivative of the

surface action potential with respect to axial distance along the excised nerve in air. Plonsey [65] in 1964, further extended the formula of Lorente de Nó, and derived equations dependent upon in situ measurements: the membrane current density and the in situ surface potential. In the following year, using the concept that the potential in a volume conductor can be expressed in terms of the solid angle subtended at any point in the field by each active area element of the membrane, Plonsey [66] extended the analysis to the case of unequal internal and external conductivities. Clark and Plonsey [18] gave a mathematical evaluation of the core conductor model of a nerve fiber put forward by Hermann in 1879 [67]. They concluded that the core conductor model is a good approximation for the internal but not the external parameters. Restricting the analysis to only the internal parameters of the core conductor model, Clark and Plonsey [19] showed that it predicted the relationship between the membrane current and the second derivative of the transmembrane potential. This relationship had been experimentally shown by Tasaki [68].

In the study of muscle fibers, Håkansson [16] examined experimentally the volume conducted action potential of an isolated frog muscle fiber. He demonstrated that the intracellular action potential had a monophasic time course and that the extracellular action potential was diphasic. He pointed out that the third positive phase which was present in the volume conducted potential of the nerve did not appear for the muscle because, he explained, of the slower course of repolarization. Rosenfalck [17] mathematically accounted for the extracellular action potential recorded by Håkansson in terms of the intracellular potential, and he further compared his findings with the core conductor model and the theory of Lorente de Nó. He also indicated that the dipole concept

presented by other investigators [69,70,71] can be used to approximate the extracellular action potential of the muscle fiber. More recently, Dimitrova [72] proposed a model of the single muscle fiber taking into account finite length, and the presence of two depolarized zones which spread in opposite directions when a nerve impulse arrives at the motor end plates [73,74].

3.1.2 Previous Single Motor Unit Models

Relatively few models of the single motor unit potential in muscle, have been described in the literature. Lindström [12] proposed a power spectrum model ofthe single fiber and in a later paper, Broman and Lindström [22] extended the expressions derived for the single muscle fiber to a mathematical model for the Fourier transform and power spectrum of a motor unit action potential. The influence of dispersion of the individual fiber signals, the spatial arrangement of those fibers and the action potential velocity, on the power spectrum of the motor unit signal, was studied. R.E. George [20] investigated the simulation of single muscle fibers using a dipole model in the time domain. He considered the summation of the single fiber potentials at a point situated at the center of a cylindrical dipole array. He also explored how small amounts of scatter in the axial position affect the spreading of the potential peaks.

3.1.3 Description of the Model

In this chapter, a model for the generation of the action potential from the single motor unit is proposed. This model incorporates known physiological parameters such as the spatial arrangement of the fibers, the velocity of the fiber action potential and the axial scatter of the single fiber potentials. A dipole model for the single fiber

potential is used as an approximation to the action potential. The form of the motor unit potential (MUP) is studied at points inside and outside the motor unit territory. The effect on the amplitude and duration of the MUP viewed from outside the motor unit, is discussed. In addition, an approximation to the multielectrode as described by Buchthal [75] is used to compare his experimental data with the results obtained from the motor unit model.

3.2 The Development of the Model

3.2.1 Single Fiber Potential Representation

In his studies on normal human single muscle fibers *in situ*, Ekstedt [76] showed that both during voluntary and chemical activation the action potential shape recorded was a clean and smooth biphasic spike. He also pointed out that for an electrode kept in a constant position relative to the fiber, each consecutive single fiber action potential had an identical shape. The dipole model was thus chosen as a first approximation to simulate the extracellular action potential of the human muscle fiber. A dipole generator may be considered as consisting of a source of current I and a sink of current I at a distance 2b apart as shown in Fig. 3.1.

A current source I would produce a spherically symmetric potential field ϕ given by

$$\phi(r) = \frac{I}{4\pi\nu r}$$

where r is the distance from the source and ν is the conductivity of the medium surrounding the fiber. Thus the potential produced at a point (x,y) (Fig. 3.1) from a dipole with source at (0,-b) and sink at (0,b) is

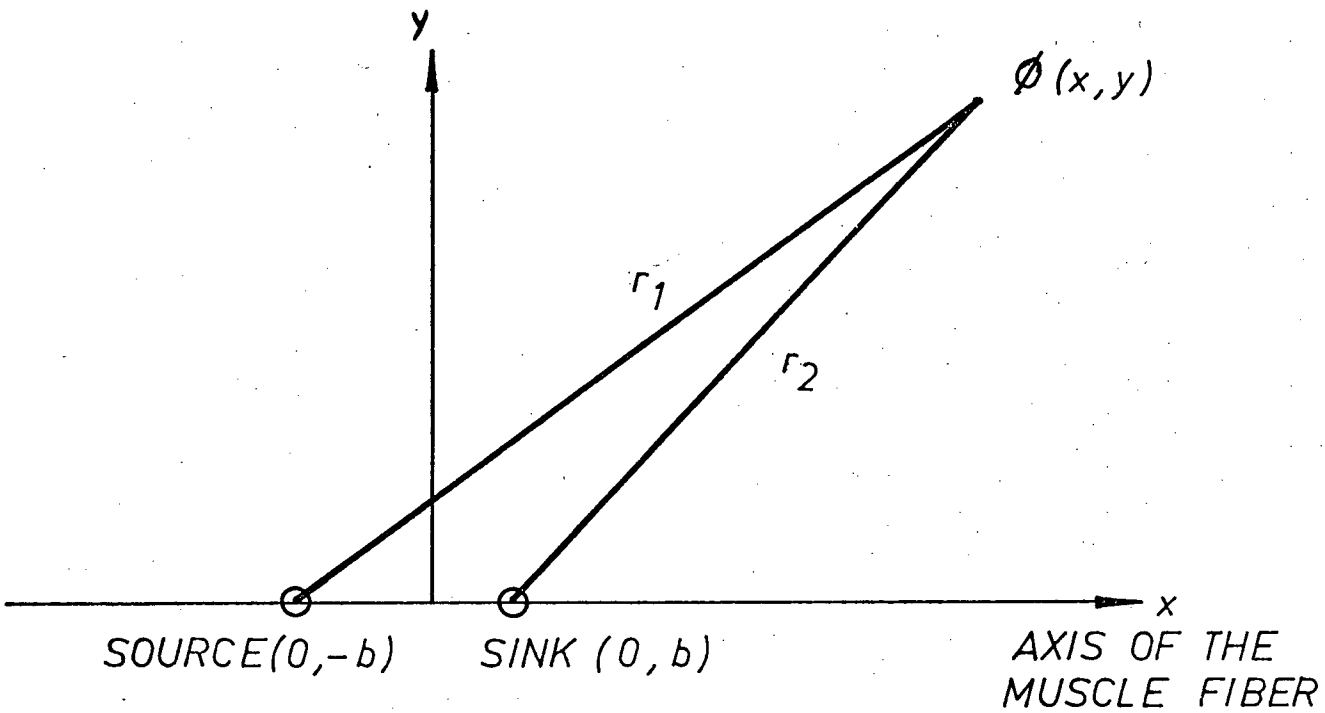


FIG. 3.1 THE SINGLE FIBER DIPOLE

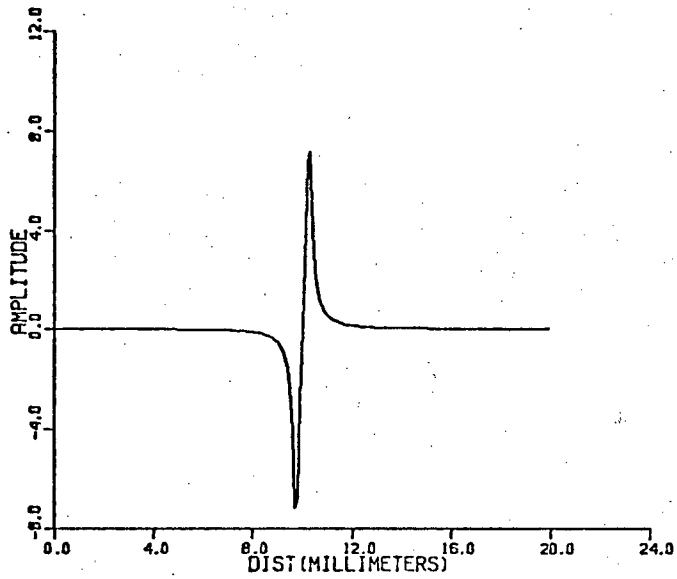
$$\phi(x,y) = \frac{I}{4\pi\nu}\left[\frac{1}{r_1} - \frac{1}{r_2}\right]$$

$$\phi(x,y) = \frac{I}{4\pi\nu}\left[\frac{1}{\sqrt{(x+b)^2 + y^2}} - \frac{1}{\sqrt{(x-b)^2 + y^2}}\right] \qquad (3.1)$$

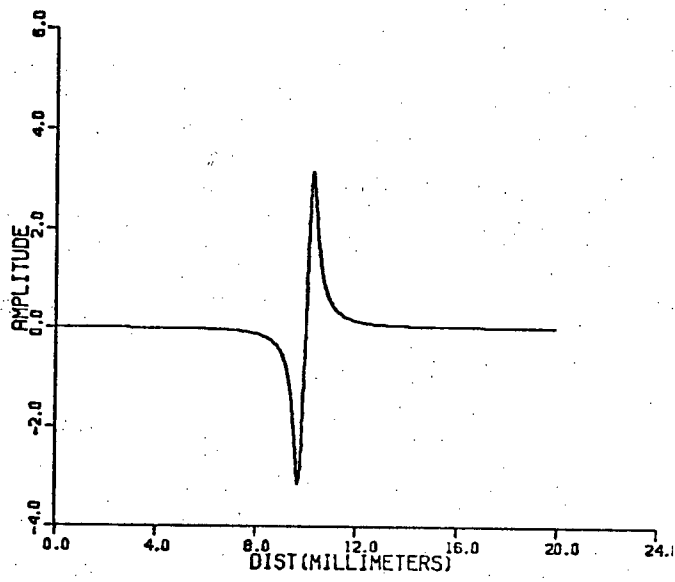
Since the motor unit structure of the biceps brachii muscle has been described in the literature, this muscle was chosen as the basis for the model. In the biceps brachii muscle, the mean propagation velocity for motor unit potentials was given by Buchthal [77] to be 4.1 m/sec (at 36.5°C) and thus the single fiber potential can be assumed to have the same velocity. The peak to peak duration close to the fiber is dependent upon the length of the dipole. The separation of the source and the sink of the dipole was chosen to be 0.5 mm (i.e. b = 0.25 mm). At the propagation velocity of 4.1 m/sec the minimum peak to peak duration was therefore 121 μsec. This result lies within the range as observed by Ekstedt [75] (see also Rosenfalck [17]). Using this dipole model, examples of the action potential computed at different distances from the fiber are shown in Fig. 3.2.

3.2.2 The Fiber Array

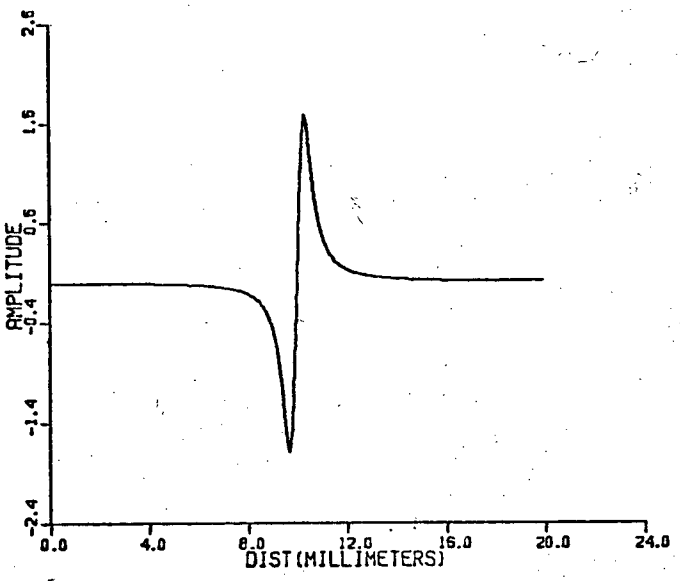
Buchthal *et al.* [78] reported that the territory of one motor unit in the normal human brachial biceps muscle is nearly circular with a mean diameter of 5 mm. It has been estimated that a motor unit in this muscle contains an average 163 fibers [73]. The spatial arrangement of the fibers in a motor unit has also been explored. In the histological studies of glycogen depleted rat fibers, Brandstater and Lambert [79] showed that the fibers are uniformly scattered throughout the motor unit area. More recently, Stålberg *et al.* [80] report that the fibers



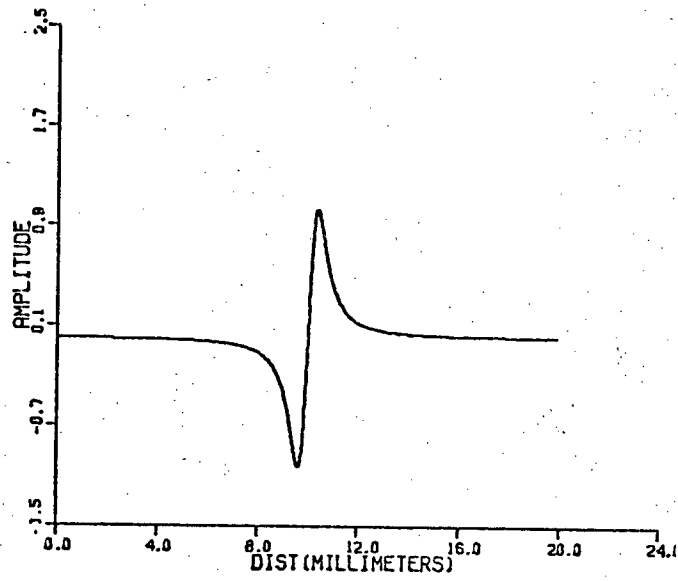
0.1 mm FROM THE DIPOLE



0.2 mm FROM THE DIPOLE



0.3 mm FROM THE DIPOLE



0.4 mm FROM THE DIPOLE

FIG. 3.2 SINGLE FIBER ACTION POTENTIAL COMPUTED AT DIFFERENT DISTANCES FROM THE DIPOLE

ORDINATE: POTENTIAL AMPLITUDE IN UNITS OF $\frac{I}{4\pi\nu}$.

ABSCISSA: DISTANCE ALONG AXIS OF THE DIPOLE IN MILLIMETERS (DISTANCE TO TIME RELATED BY CONDUCTION VELOCITY OF 4.1 m/s)

in the human bicep brachii have a similar scatter with no tendency for grouping. To simulate this spatial scatter of the fibers in the motor unit, an array of 163 fibers was computer generated to lie with a uniform random distribution within a circle of 5 mm diameter. Fig. 3.3 shows a representative array. Each muscle fiber is represented by a circle of diameter 50 μm, which is the average fiber diameter in the biceps brachii [81]. It was observed from this random array that the percentage of adjacent fibers was similar to the experimental findings of Brandstater and Lambert.

3.2.3 The Motor Unit Potential

Assuming that the fiber array lies within an extensive homogenous volume conductor, then Helmholtz' Principle [82] that the voltage at a point caused by more than one source of electromotive force (emf) is given by the algebraic sum of the voltages caused by each emf acting alone, can be applied. A FORTRAN computer program to run on the IBM 370/168 has been written which can input the coordinates of a point electrode placed inside or outside the fiber array and sum the potential contributed by each single muscle fiber to produce the motor unit potential at that point. An example of a MUP computed at the center of the array of Fig. 3.3 (at (2.5, 2.5)) is shown in Fig. 3.4. It is observed that not only the amplitude of this potential has increased over the single fiber as anticipated, but also the total duration.

3.3 The Motor Unit Potential Amplitude and Duration

Motor unit potentials for a set of points at increasing distance from the array center were generated for 15 individual random muscle fiber arrays. The maximum, minimum and mean of the peak to peak amplitude and peak to peak duration at each point were then calculated and plotted as

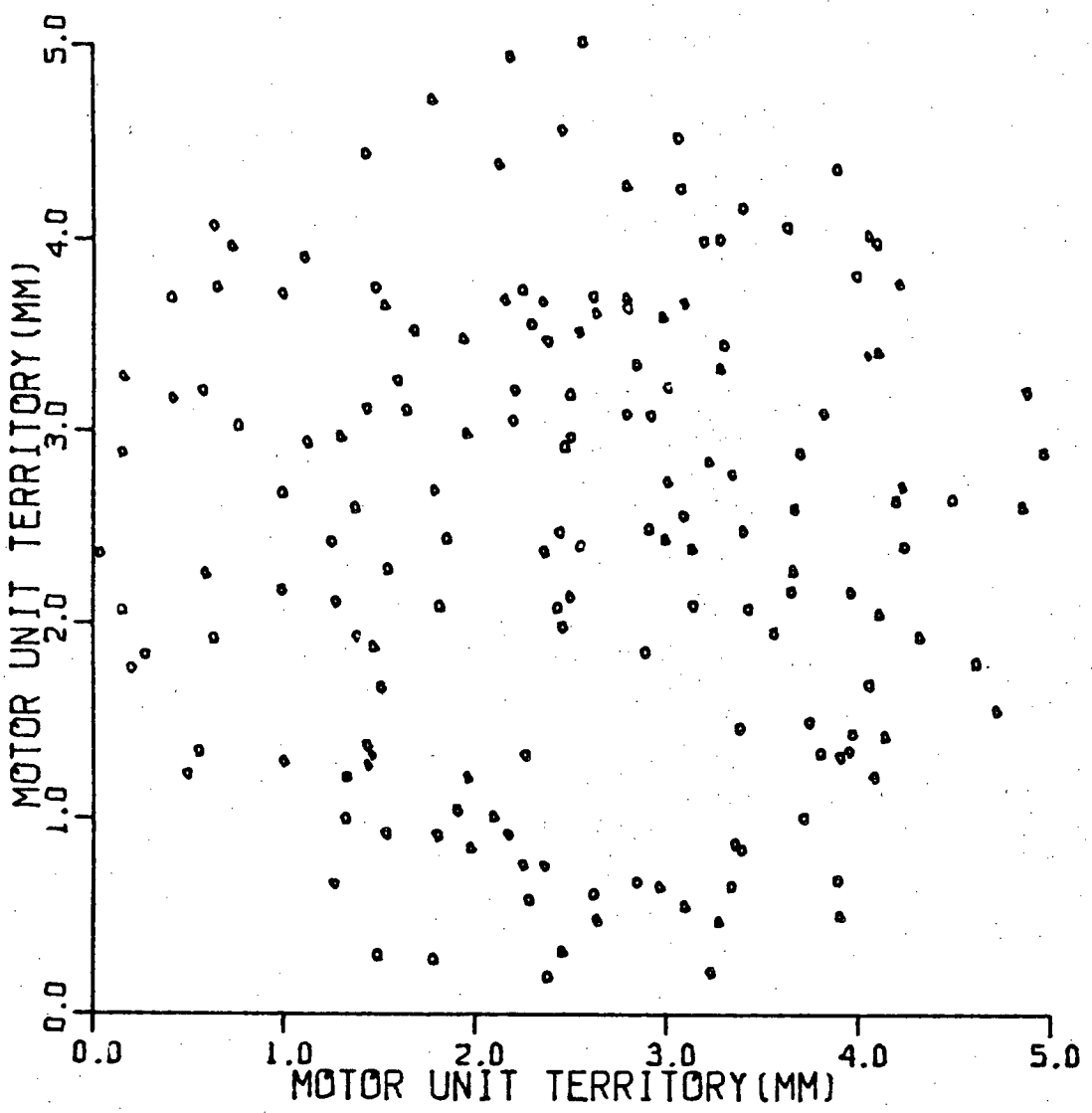


FIG. 3.3 TYPICAL RANDOM ARRAY REPRESENTING A MOTOR UNIT IN THE HUMAN BICEPS BRACHII MUSCLE (163 MUSCLE FIBERS)

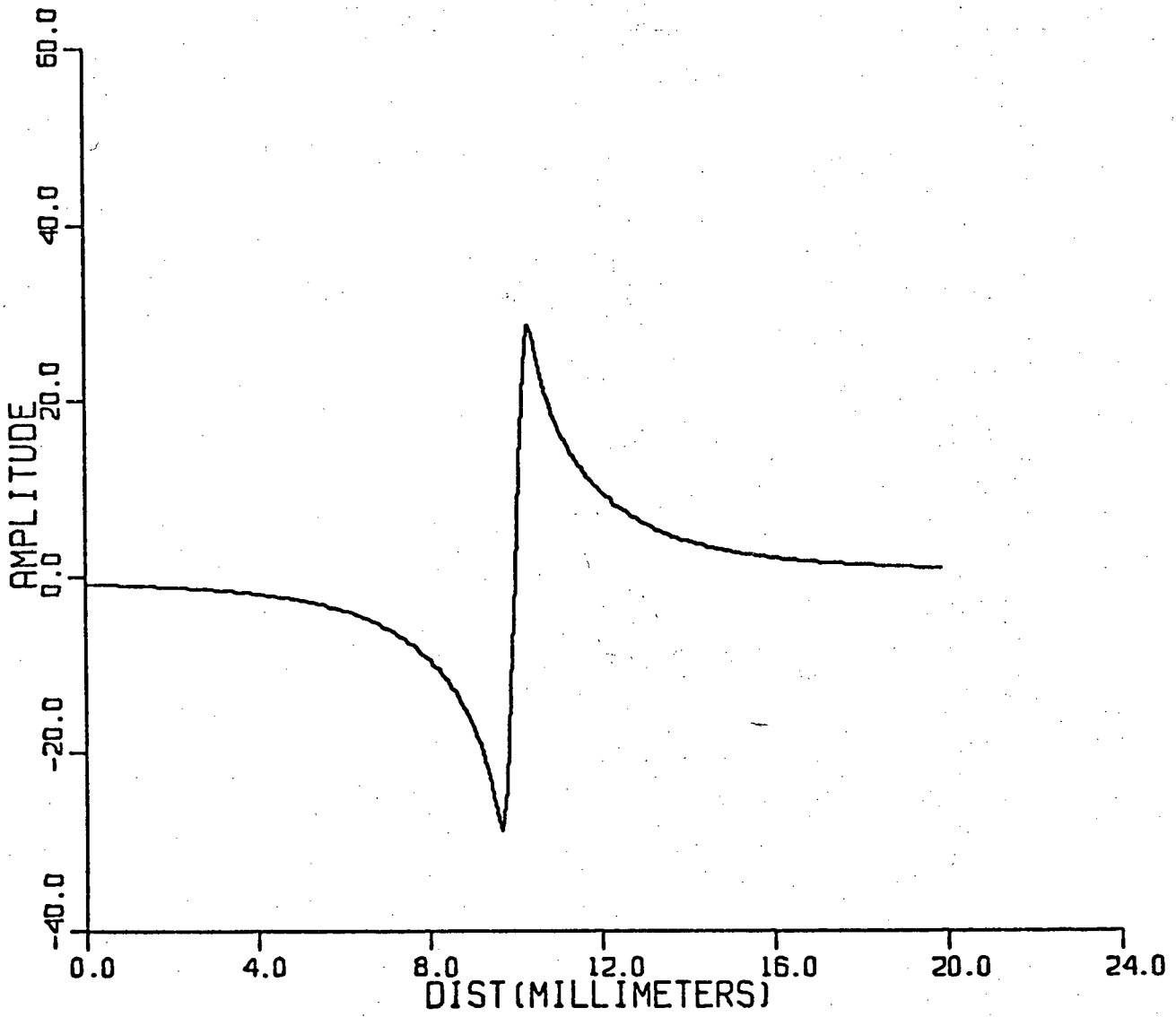


FIG. 3.4 A MOTOR UNIT POTENTIAL - COMPUTED AT (2.5,2.5) IN THE ARRAY SHOWN IN FIG. 3.3. SCALING AS IN FIG. 3.2.

a function of distance (Fig. 3.5 and Fig. 3.6). Due to the great variability inside the motor unit caused by the close proximity of the fibers, only points outside the motor unit territory are shown in these figures.

The relationship between the mean single motor unit potential peak to peak duration and the single fiber peak to peak duration is shown in Fig. 3.5. Consider one single fiber placed at the center of the array. By differentiating equation (3.1) with respect to x and setting:

$$\frac{d\phi}{dx} = 0$$

it can be shown (George [20]) that when $x >> b$, the position of the peaks of the single fiber potential is approximated by:

$$x = \pm \frac{y}{\sqrt{2}} \tag{3.2}$$

where y is the distance from the dipole and x lies on the axis of the dipole. The peak to peak duration of the potential ($2y/\sqrt{2}V$) has been plotted as a function of distance outside the motor unit territory in Fig. 3.5. Near the edge of the motor unit territory, the duration of the motor unit potential is decreased from the single fiber potential due to the influence of nearby fibers.

A comparison between the peak to peak amplitude of a single fiber at the center of the array and the mean single motor unit peak to peak amplitude can also be made (Fig. 3.6). By substitution of equation (3.2) into equation (3.1), and multiplying the result by 2 x 163, the peak to peak amplitude as a function of distance extraterritorially from the motor unit can be calculated. It can be seen that peak to peak amplitude is increased from the fiber array near the edge of the motor unit due to greater influence of fibers at the boundary.

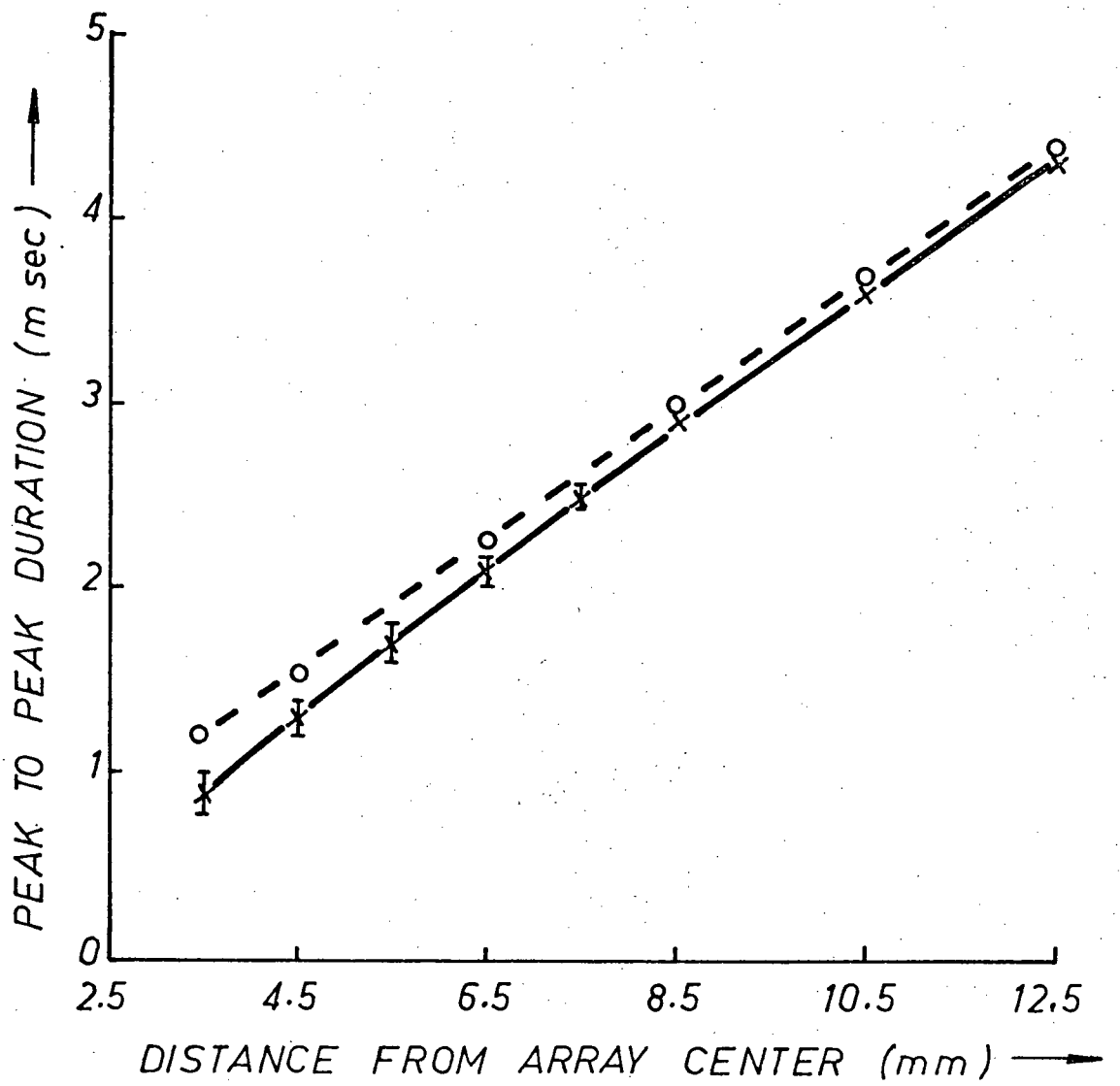


FIG. 3.5 PEAK TO PEAK DURATION OUTSIDE THE MOTOR UNIT AS A FUNCTION OF DISTANCE

- - - - SINGLE FIBER POTENTIAL

——— MOTOR UNIT POTENTIAL

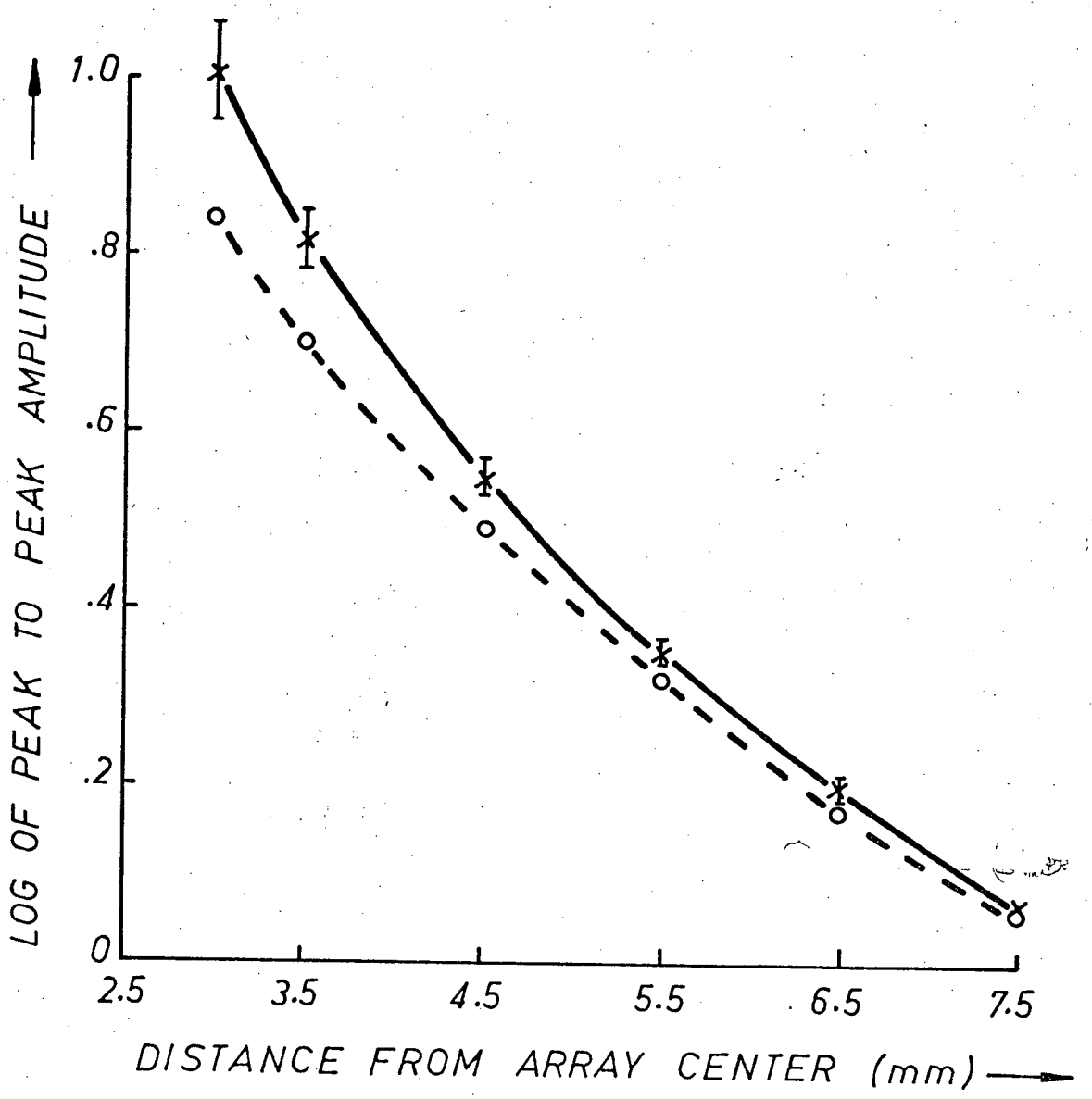


FIG. 3.6 PEAK TO PEAK AMPLITUDE OUTSIDE THE MOTOR UNIT AS A FUNCTION OF DISTANCE.
(AMPLITUDE IN ARBITRARY UNITS)

- - - - SINGLE FIBER POTENTIAL

——— MOTOR UNIT POTENTIAL

3.4 Axial Dipole Dispersion

At any instant of time, the activity centers of all the fibers in a single motor unit do not lie in the same plane normal to the direction of propagation. This is due to differences in conduction time along branching motor nerve fibers, differences in the axial position of the motor end plates they innervate [74] and finally, differences in conduction velocity of the muscle fibers within the motor unit. The result of these three factors is a temporal dispersion of arrival times of the fiber dipoles at the plane of electrode. This is indicated in Fig. 3.7.

To study the effects of dispersion on the motor unit potential, a Gaussian distribution of the fiber dipole locations is assumed. Fig. 3.8 shows the motor unit potentials calculated at distances from the center of the fiber array (2.5, 2.5 mm), and compares the application of a normal distribution of dipole positions (standard deviation of 0.7 mm) with no dipole scattering. It is to be noted, that the sampling electrode is assumed to be placed outside the zone of innervation. The effect of the dispersion is to elongate the potential and decrease its amplitude [20]. A further outcome of this distribution is to introduce irregularities in the form of spikes and notches within the motor unit. Outside the motor unit, however, the potentials become smooth and diphasic. Fig. 3.9 shows the peak to peak amplitude as a function of distance outside the motor unit territory for different values of standard deviation.

The effect of axial scatter of the fiber dipoles on the peak to peak duration of the motor unit potential is shown in Fig. 3.10. Increasing the standard deviation of the fiber dipole distribution, results in irregularities in the motor unit potential and eventual splitting of the potential. This is shown in Fig. 3.11.

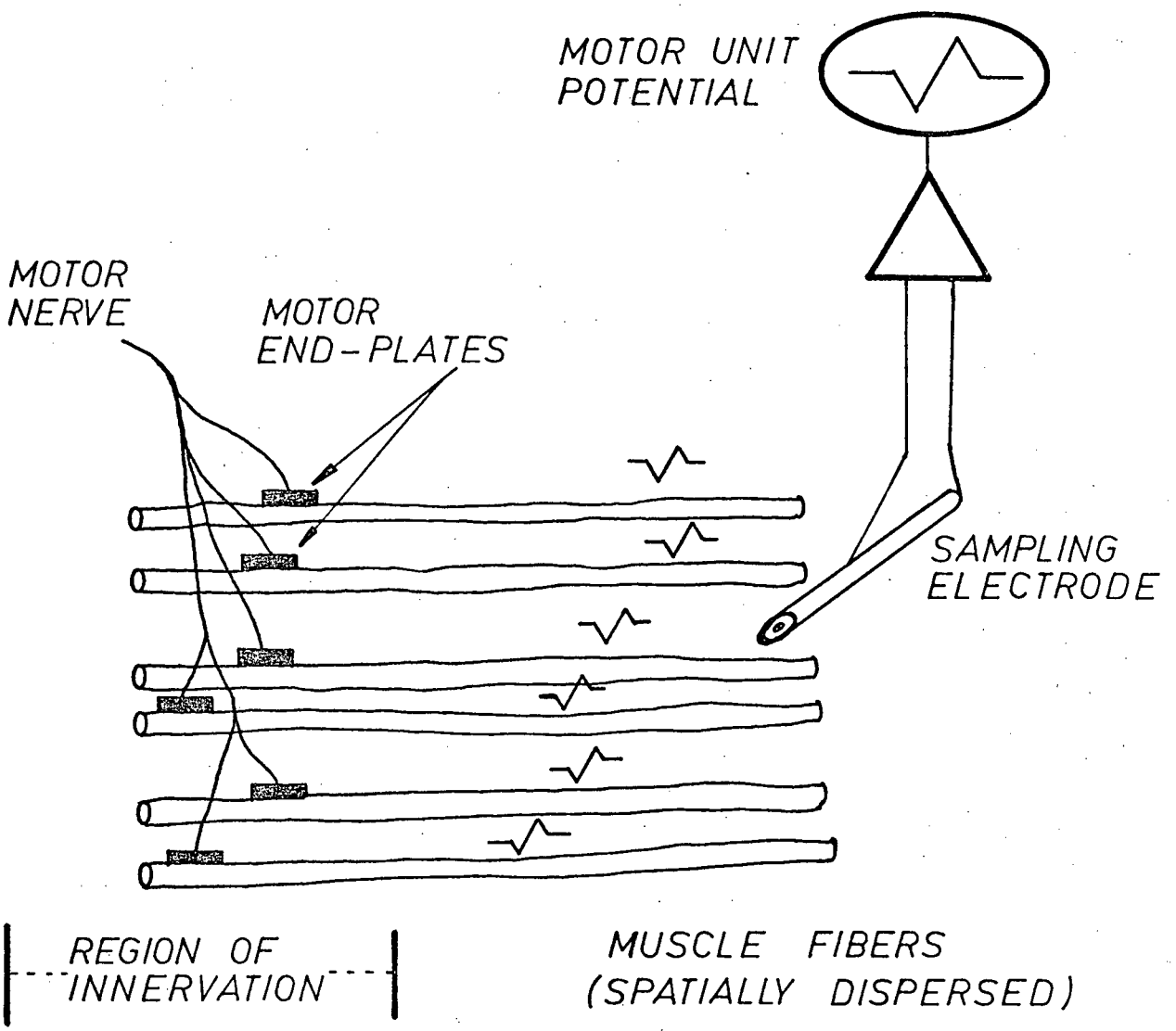


FIG. 3.7 DIAGRAMMATIC REPRESENTATION OF THE MOTOR UNIT ANATOMY CONTRIBUTING TO THE DISPERSION OF THE SINGLE FIBER POTENTIALS

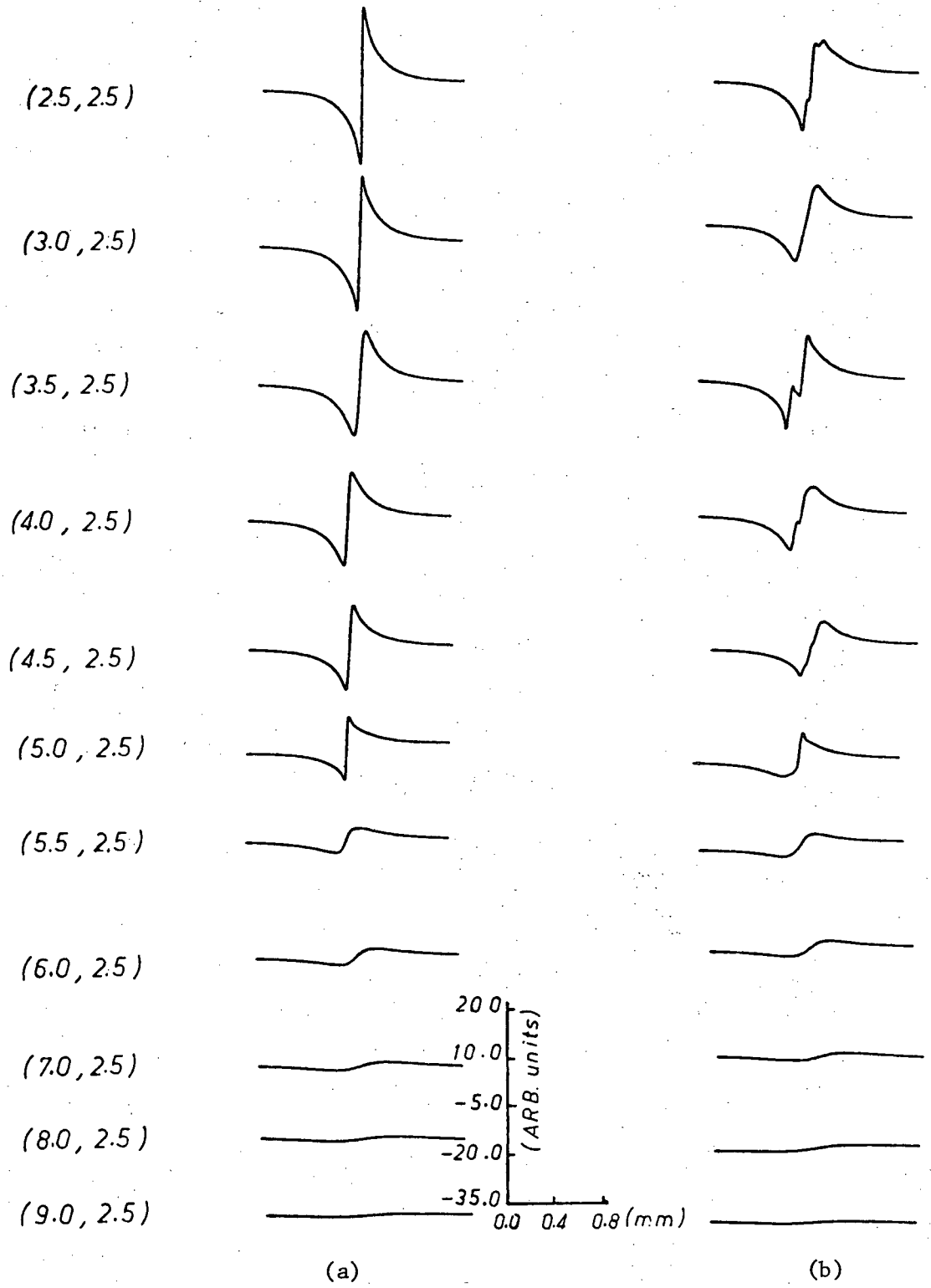


FIG. 3.8 EFFECT OF DIPOLE SCATTER ON THE SHAPE OF THE MOTOR UNIT POTENTIAL
(a) NO SCATTER
(b) GAUSSIAN SCATTER WITH STANDARD DEVIATION = 0.7 mm.

CO-ORDINATES REFER TO THE SAMPLING POSITION WITH RESPECT TO THE FIBER ARRAY.

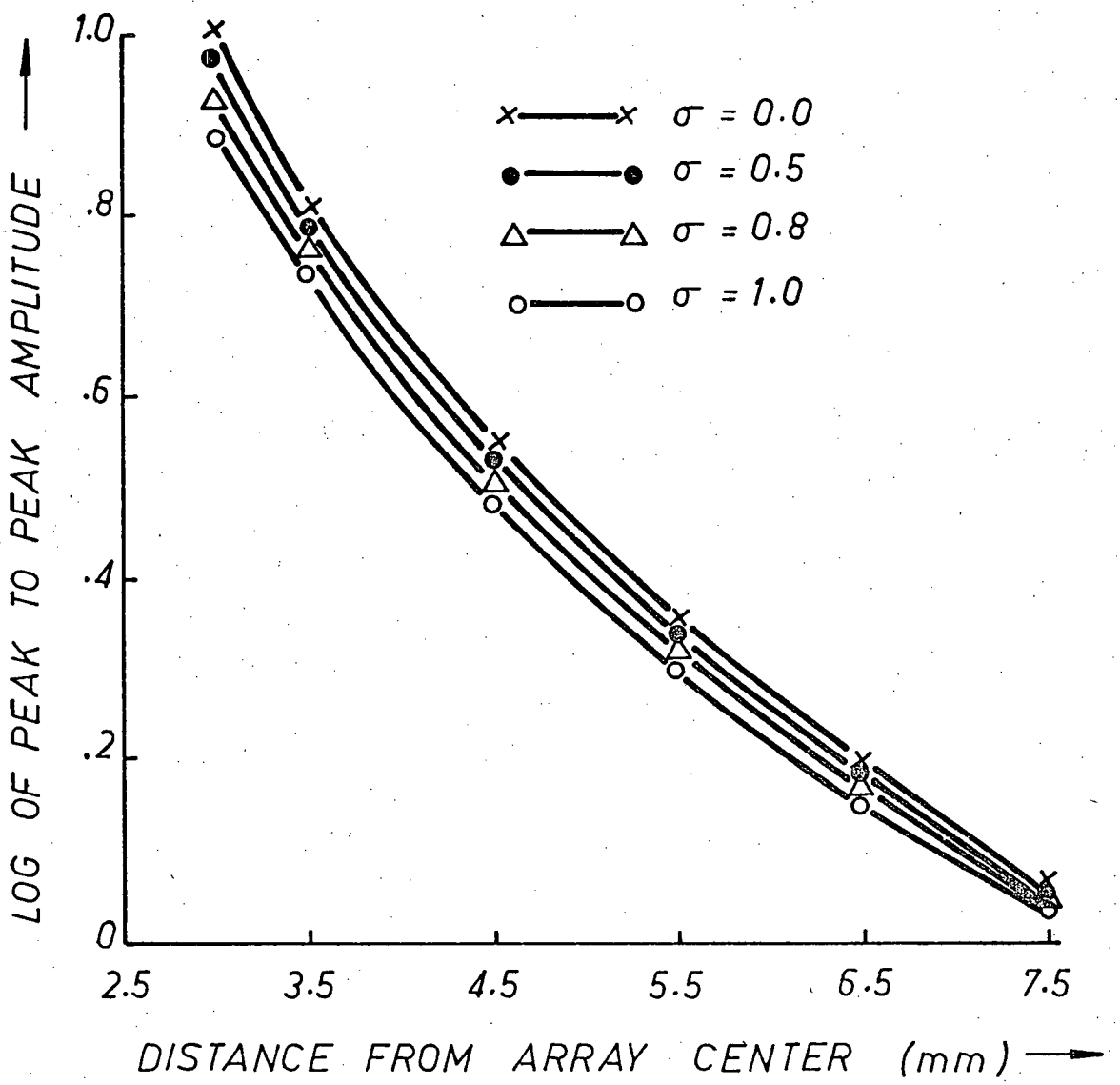


FIG. 3.9 EFFECT OF INCREASING THE DISPERSION OF DIPOLES ON PEAK TO PEAK AMPLITUDE
(AMPLITUDE IN ARBITRARY UNITS)

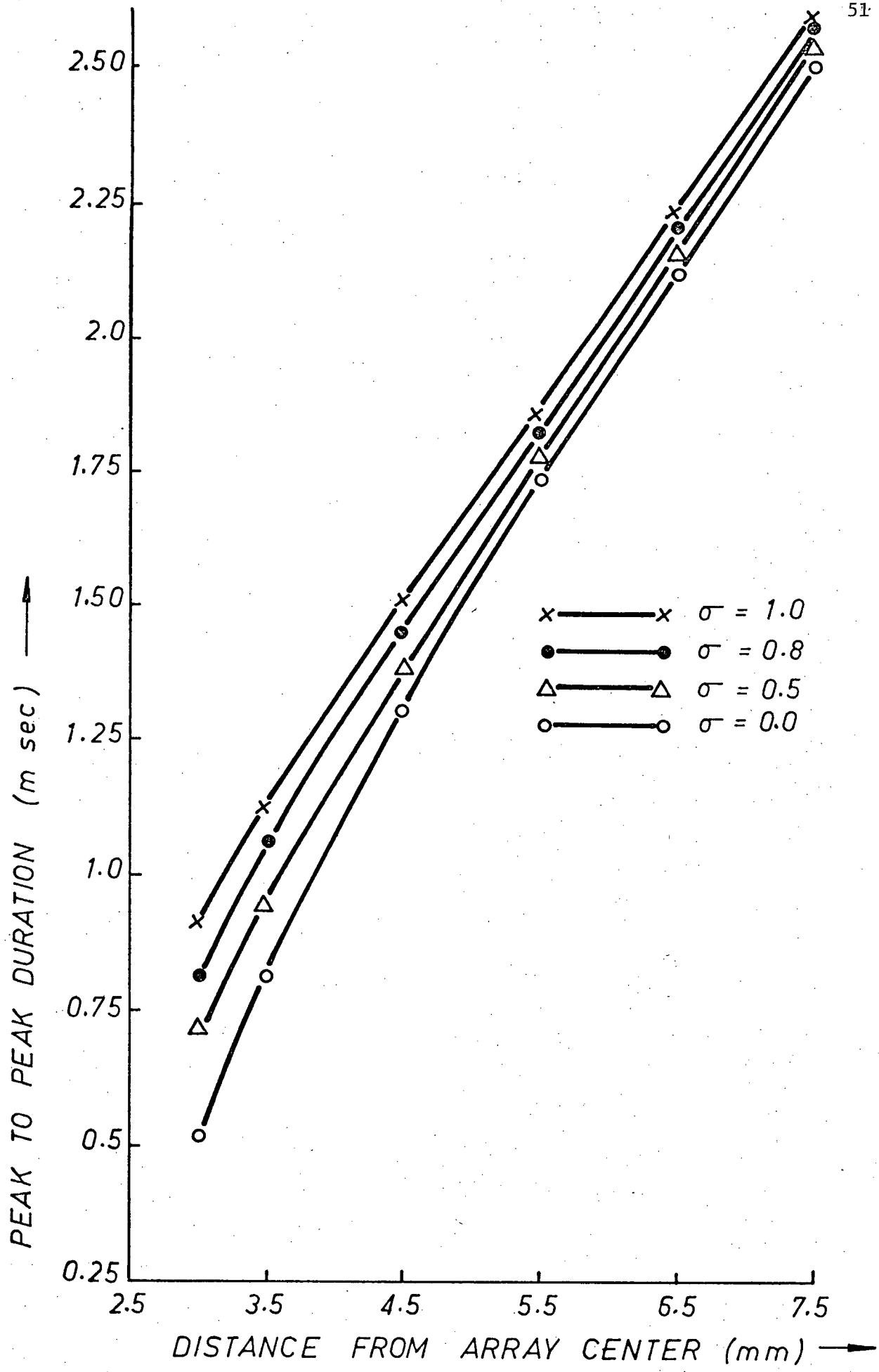
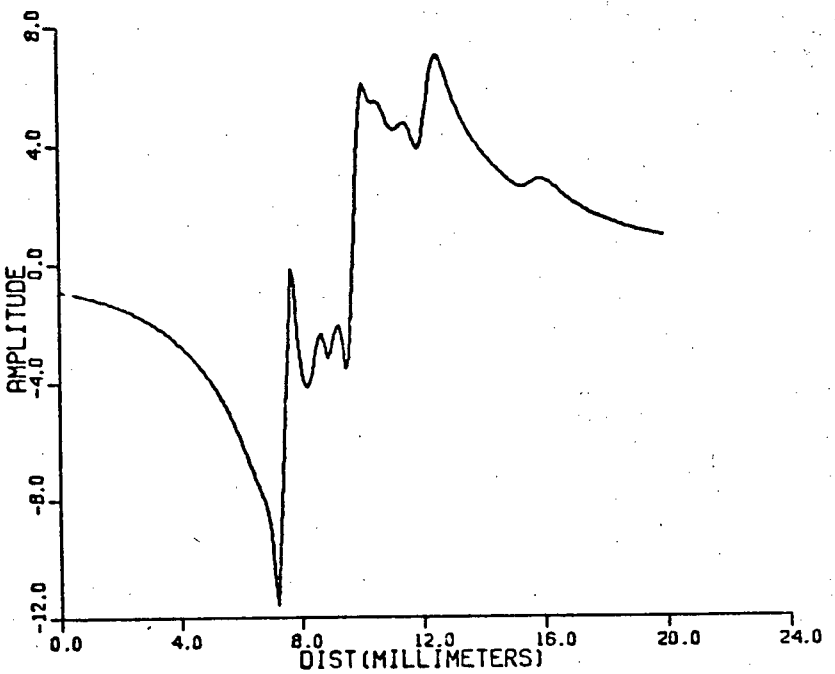
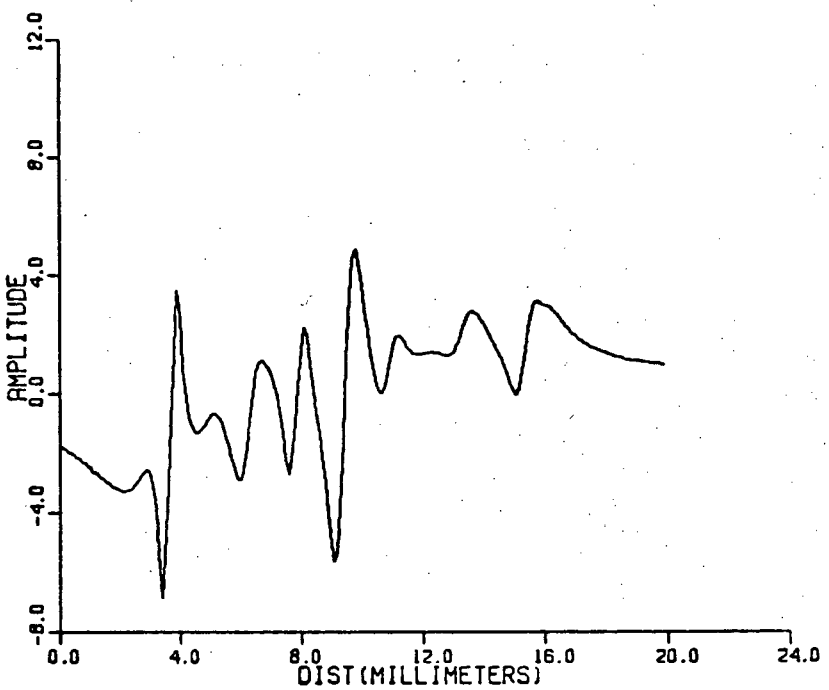


FIG. 3.10 EFFECT OF INCREASING THE DISPERSION OF THE DIPOLES ON PEAK TO PEAK DURATION



GAUSSIAN DISPERSION OF THE DIPOLES WITH STANDARD DEVIATION = 2.0 mm



GAUSSIAN DISPERSION OF THE DIPOLES WITH STANDARD DEVIATION = 5.0 mm

FIG. 3.11 EFFECTS ON THE MOTOR UNIT POTENTIAL SHAPE FOR LARGE VALUES OF AXIAL DIPOLE DISPERSION

ORDINATE: AMPLITUDE IN ARBITRARY UNITS.

ABSCISSA: DISTANCE IN MILLIMETERS (CONDUCTION VELOCITY=4.1 m/s).

MOTOR UNIT POTENTIALS COMPUTED AT CENTER OF MUSCLE FIBER ARRAY (2.5,2.5)

3.5 Comparison With Experimental Data

Buchthal has explored the territory of the motor unit in the biceps brachii by means of a multielectrode [83,75]. He describes this electrode [75] as having twelve 1.5 mm long leads, each placed 0.5 mm apart over a length of 25 mm. The electrode was inserted at right angles to the longitudinal axis of the muscle fibers and the peak to peak amplitude through the motor unit area was plotted against distance. His results [84, Fig. 3] outside the motor unit territory are compared with those obtained from the model for several dispersions of the motor endplates. These plots are shown in Fig. 3.12. Each graph has been normalized to the peak to peak amplitude at 7.5 mm from the array center. If it is assumed that the voltage led off from any electrode surface is equal to the average voltage in the volume conductor over the recording area, then the multielectrode may be better simulated by taking potential values at points and averaging the computations over the area of the electrode. Such a technique has been employed by Ekstedt and Stålberg [21] and was originally proposed by Håkansson [85]. Using the dimensions given for the multielectrode, an average of 15 points was taken for each recording surface and the peak to peak amplitude was plotted against distance through the motor unit territory and compared with the findings of Buchthal in Fig. 3.13. The plots have been normalized to the center of the motor unit territory where the motor unit potential had the largest amplitude.

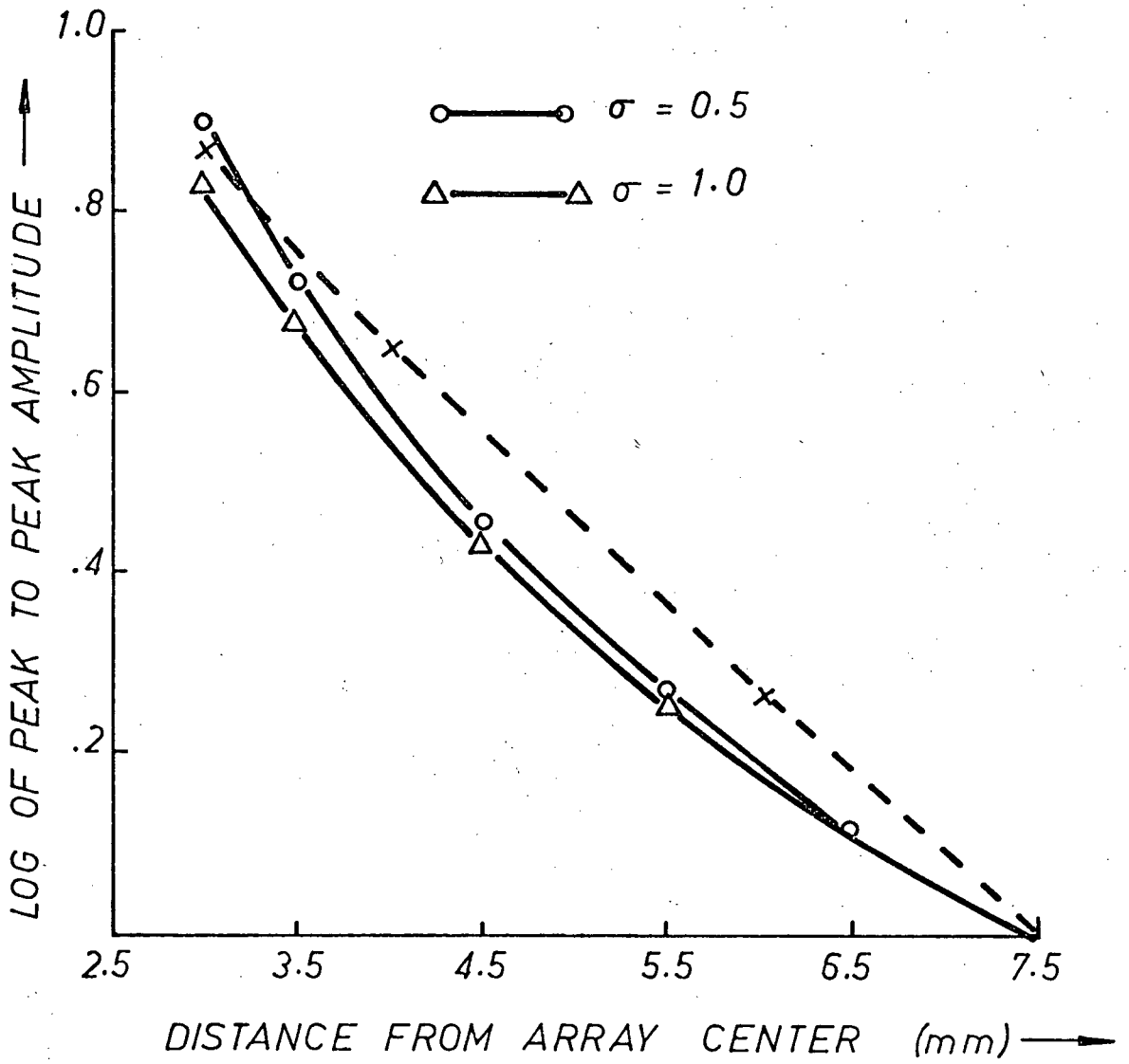


FIG. 3.12 COMPARISON WITH EXPERIMENTS OF BUCHTHAL [84]

- - - - EXPERIMENTAL DATA

——— POTENTIAL AT A POINT

(PLOTS ARE NORMALIZED TO PEAK TO PEAK AMPLITUDE AT 7.5 mm FROM ARRAY CENTER.)

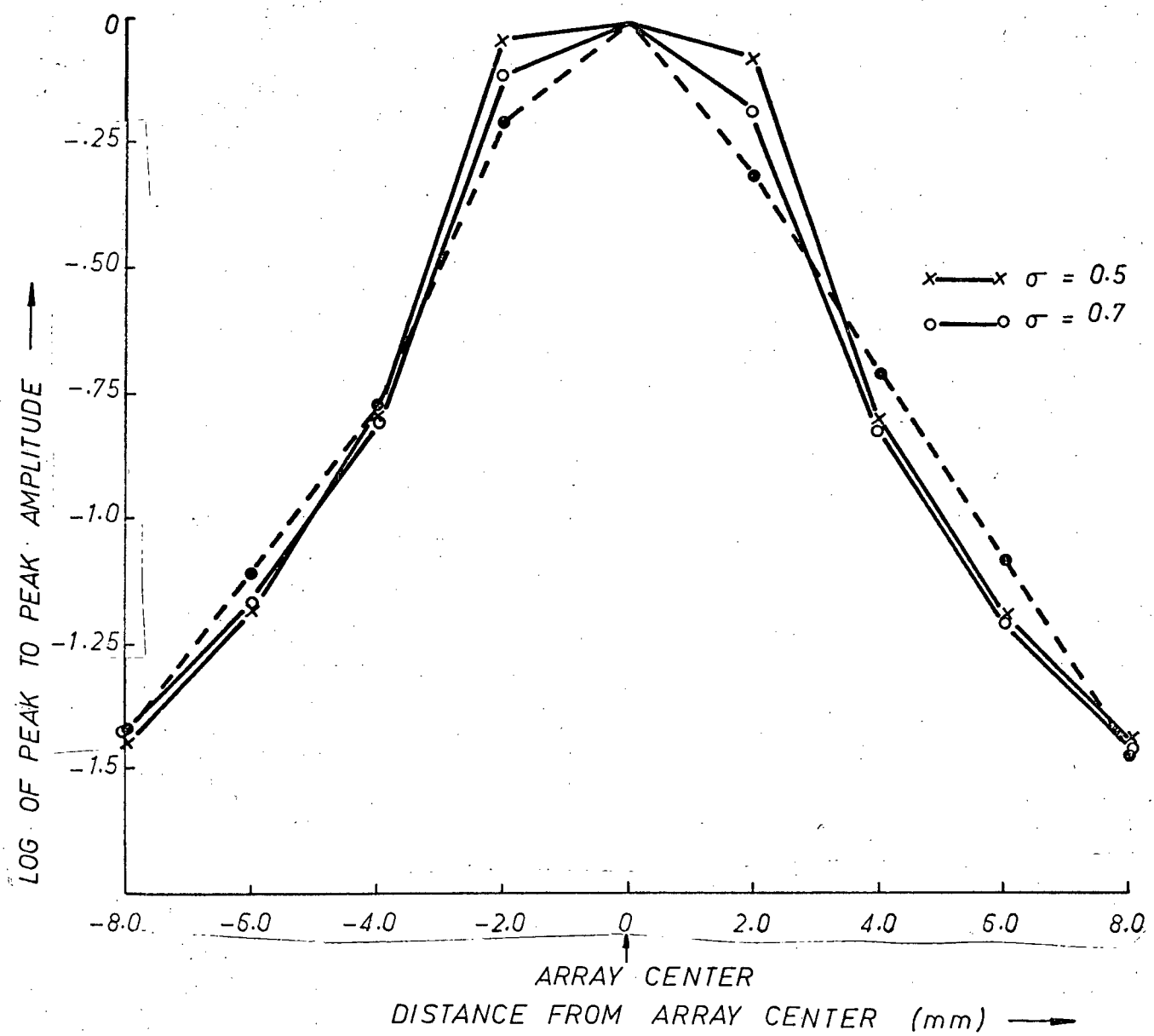


FIG. 3.13 COMPARISON WITH EXPERIMENTS OF BUCHTHAL [84]

- - - - EXPERIMENTAL DATA

——— SIMULATION OF THE MULTIELECTRODE

(PLOTS ARE NORMALIZED TO THE MAXIMUM PEAK TO PEAK AMPLITUDE AT THE CENTER OF THE ARRAY)

3.6 Discussion

3.6.1 Validity of the Model

The biphasic form of the single fiber action potential is represented by a longitudinally orientated dipole, travelling along its axis. Thus, at any time, one part of the muscle fiber is considered as a source of current giving rise to an emf, while another part is acting as a sink. The action potential is a transient disturbance of the resting state such that the sources and sink must be equal. In this simple representation, the finite diameter of the muscle fiber is not considered. Other considerations which have not been incorporated into the model are the effects of inhomogeneity, anisotropy, the effects of other active fibers in the vicinity and the effects of the fiber potentials propagating in the other direction from the innervation points. Since the form of the fiber potential is of similar shape as recorded by Ekstedt [76], the author considers these effects to be second order.

The territory of the motor unit in the biceps brachii has been estimated to be circular with an average diameter of 5 mm. Stålberg *et al.* [80] using single fiber electromyography, have recently reported that the muscle fibers of one motor unit are scattered over a distance of less than 8 mm in the biceps brachii. The fiber array used in this model is based on the mean value given for the motor unit territory, although variations from 2-14 mm have been reported [86].

Histological mapping of the motor unit in man has not been made. The number of muscle fibers within a motor unit is thus based on an estimate. Buchthal and Madsen [87] in 1950, first suggested that there was an average of 1000 muscle fibers in a motor unit of the biceps brachii muscle. They based this on an estimate of the total number of muscle

fibers in man and the total number of myelinated nerve fibers, giving an average of 700 fibers per motor unit in the human body. They conclude that, "it seems therefore, reasonable in a large muscle like the brachial biceps to reckon with an average of 1000 fibers per motor unit". The number of 163 fibers chosen for the model is based on the histological examinations of Christensen [73] and is calculated from counting the total number of neurofibrils and the total number of muscle fibers of the biceps brachii muscle. Using the investigations of Feinstein et al. [88], 60 per cent of these neurofibrils are considered as motor nerves and hence from the total number of muscle fibers, the number of fibers per motor unit can be found. There is no correction however, for the small motor nerve fibers supplying the intrafusal muscle fibers of the muscle spindles [89] and for the fact that multiple innervation may occur [74]. It should be further pointed out that no indication of the variability between different subjects is given. It is also unlikely that every motor unit in the biceps brachii has the same number of fibers.

The velocity of the action potential was chosen as 4.1 m/sec, which is based on the results of the experiments by Buchthal et al. [81] on electrically evoked action potentials. Using this value, the summated motor unit potentials calculated from the fiber array were found to be within the total duration range reported by Buchthal [10]. This range is from 2-20 msec total duration in the normal motor unit potential recorded in the biceps brachii muscle. Short durations are found near and within the motor unit and long durations further away.

In comparing the peak to peak amplitude as a function of distance through the motor unit territory with experimental findings, a similarity was observed for both the potential at a point and with the

simulation of the multielectrode. Correction for the effect on the volume conducted field of introducing an electrode of relatively large dimensions, such as the multielectrode has not been considered in the simulation.

3.6.2 Implications of the Model

The motor unit potential calculated from the uniform randomly distributed muscle fiber array has a diphasic form (Fig. 3.3) and corresponds to normal findings in clinical electromyography. The summation of the contribution from the individual muscle fibers shows clearly and explains the finding in electromyography that the total duration of the motor unit potential is longer than the fiber action potential duration. The effect of volume conduction on this summation has been demonstrated by plotting the peak to peak duration and amplitude as a function of distance. The results show that the duration increases with distance while amplitude decreases. Bauwens [90] described an experiment in which ten distinct diphasic potentials of 1 msec total duration were generated and summated electronically. As the potentials were merged together, they formed a diphasic potential of greater amplitude and duration. He suggests however, that the blending of these potentials cannot account for a duration of more than 3 milliseconds, and the explanation for longer durations is related to the electrical characteristics of the tissue and also the position of the electrode. The model described here accounts for total durations larger than 3 milliseconds, because potentials from the dipoles at a distance from the computation point contribute to the low amplitude initial and final phases of the motor unit potential. The position of the electrode outside the motor unit territory increases duration confirming Bauwen's prediction.

A Gaussian scatter of the dipoles along the fiber axis was applied to simulate the differences of the arrival at the electrode of the single fiber action potentials. This accounts for the combined effects of the motor end-plate dispersion, any conduction velocity differences of the muscle fibers and differences in the conduction time along the terminal nerve endings from the motor nerve to the end-plate. The results from the model have shown that the duration of the motor unit potential increased with increasing dispersion and that the amplitude decreased.

In the biceps brachii there is a three-fold variation in muscle fiber diameter, and the investigations of Håkannson [85] on an isolated frog muscle fiber in Ringer's solution have shown that the conduction velocity was a linear function of the fiber circumference. Despite these facts, the conduction of the human brachial biceps only varies between 4 - 5.5 meters/sec. This small variation may be accounted for by the influence of other muscle fibers *in situ*. Buchthal *et al.* [77] did not consider this difference in propagation velocities of the muscle fibers important in the exploration of the dispersion in arrival times at the electrode of the single fiber potentials. They also said that the propagation times in the terminal nerve did not contribute to the action potential duration. Under these assumptions, the nerve impulse would arrive almost synchronously at all the end-plates in a motor unit. This opinion has been shared by other investigators [91,6]. Buchthal *et al.* [77], who recorded the action potentials within the innervation zone found there was an almost simultaneous initial deflection to within 0.5 msec for a distance of up to 40 mm, when recording from the same motor unit. Thus, they concluded this distance is the extent of the end-plates for a motor unit. If the basic assumptions of these investigators are

correct, then on the basis of the model described here the extent of the motor end-plates cannot be greater than 10 mm otherwise the motor unit potential will be severely split (Fig. 3.11), such that it does not compare with clinical findings. This smaller predicted value for the extent of end-plates in one motor unit compares with histological findings in infants [74]. Increasing the number of dipoles within the array to 1000 results in an action potential of the same form, and therefore predicts a similar value.

CHAPTER IV

AN INVESTIGATION OF THE VARIATION IN SHAPE OF THE MOTOR UNIT POTENTIAL

4.1 Introduction

4.1.1 Motivation

Clinical observations of EMG data have indicated that in succesive firings of the single motor unit potential, there is a variation in its shape [76]. It has also been observed that, in certain diseased states of the muscle, this variability of the motor unit potential increased [59-P169,62]. To the knowledge of the author, there has been no investigation of this variation in shape or of the factors that may contribute to it.

One likely factor that may vary the shape of the motor unit potential at each discharge, is electromyographic jitter. This electromyographic jitter is the variability in the time interval between two action potentials, from two muscle fibers of the same motor unit, at consecutive discharges [92]. Thus any small variation about a mean time interval between the single fiber potential firings would give rise to summated motor unit potentials each of a slightly different shape. At present, the study of the jitter phenomena is achieved by means of a special electrode (single fiber electrode) which can record single muscle fiber potentials. In standard electromyographic procedures however, a concentric needle electrode is used, which samples the summated activity from many muscle fibers. An outcome of an investigation of the variation in the shape of the motor unit potential, if jitter can be attributed to its cause, is that it may be possible to examine the effects of jitter without the use of the single fiber electrode. Recently, electromyo-

graphic jitter has proved to be of use in the diagnosis of disease, particularly in disorders of neuromuscular transmission, such as myasthenia gravis. If the jitter changes in the diseased state, this effect may become evident in the motor unit potential variability.

In different types of diseases, there may be other physiological mechanisms that may also change the shape of motor unit potential on successive discharges. Investigation of the pathological processes is necessary before the shape of the potential in the diseased state can be fully understood. If the variability in motor unit potential shape can be proven to be an indicator of abnormality, this would become a useful addition to an automatic analysing system to aid the clinician in his diagnosis.

With the above motivations, and the tools developed in the preceeding chapters, the variation in the shape of the motor unit potential on consecutive firings, was investigated.

4.1.2 The Jitter Phenomenon

In 1964, Ekstedt [76] observed that when he recorded the action potentials from two muscle fibers from the same motor unit there was always a variability in the time intervals between the two potentials. This variability he termed "jitter". His recordings were performed with a special type of multielectrode [93], with up to 14 leading-off surfaces of 25 μm^2, adjacent electrodes being placed 60 μm apart. These dimensions are of the same order as the muscle fiber diameter (40-80 μm). When the multielectrode is inserted into the muscle it is possible to have two muscle fibers from the same motor unit so close that their potentials could be picked up with one electrode, or if the fibers were more separated, from the leading off surfaces closest to the fibers. If the

sweep of an oscilloscope was triggered by the first potential, then the second potential of a pair for each consecutive discharge appeared to move or "jitter" about the screen (Fig. 4.1).

As mentioned in section 3.4, the single muscle fiber potentials arrive at the electrode with a mutual time difference (Fig 3.7). The reasons for this may be: (1) different propagation times in the terminal nerve endings, (2) the synaptic delay of the motor end-plates may be different, (3) the position of the motor end-plates on the muscle would give rise to an unequal distance for the potentials to travel to the electrode, and finally (4) the propagation velocity of the two fibers may be different. However, the jitter phenomenom could only be explained by a variability in each successive discharge of any of the above factors. Provided there is a steady contraction of the muscle, Stålberg et al. [92] have shown that a variability in the normal muscle of the propagation velocity is not an important factor. By analogy, Ekstedt and Stålberg [94] assume that the variability in propagation velocity of the terminal nerve endings is probably not important. Since the distance from the nerve to the sampling electrode remains the same, it is reasonable that there is a variability in the synaptic delay of the normal end-plate. It has also been shown by Ekstedt and Stålberg [95] that injection of D-tubocurarine, which only affects the motor end-plate transmission, increases the jitter.

Jitter is found to be increased in myasthenia gravis although normal values of jitter may be found [96]. In more severe cases of this disorder, the jitter increases further and fewer normal values are found, but there is an occasional misfiring or blocking of the single fiber action potential. In these cases the jitter phenomenon is more difficult to measure. In the less severely affected cases of myasthenia

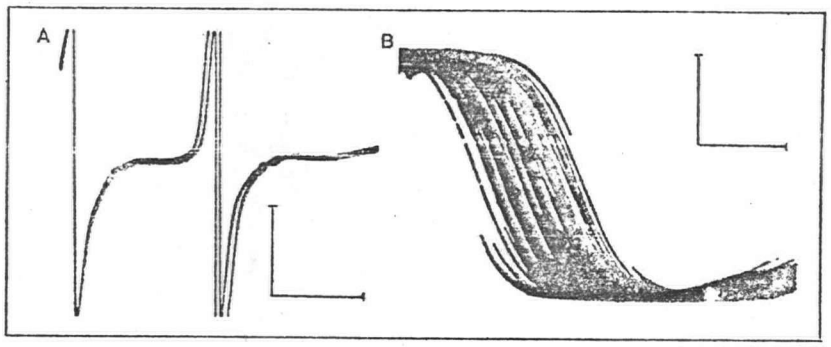


FIG. 4.1 THE JITTER IN A POTENTIAL PAIR IN THE BICEPS BRACHII MUSCLE.

A. Two discharges of the pair. The sweep is triggered by the first action potential and the second appears with different time delays in the two discharges. Calibration: 4 MV and 1,000 μsec.

B. The second action potential in the pair. About 400 discharges are superimposed. The first potential in the pair is constant on the screen. The second does not have a constant position due to the jitter phenomenon.

[From Stålberg et al, 1971]

gravis the measure of jitter is a sensitive diagnostic indicator [97,98].

Stålberg and Ekstedt [96] have shown that in muscular dystrophy, there is an increase of the jitter in 10-15% of the recordings. They suggest this may be due to changes in the nerve twigs and/or the motor end-plates. They also report findings of markedly increased jitter in neurogenic disorders and propose that this may be due to disturbed transmission in the nerve endings and to unreliable transmission across the neuromuscular juction.

4.2 Simulation of Electromyographic Jitter

As an initial step towards the study of variation in the motor unit potential shape at consecutive discharges, the computer simulation model described in Chapter III was modified to include electromyographic jitter of the single fiber potentials. An ensemble of motor unit potentials were generated and the variance as a function of time computed. By this method, the contribution of jitter to the variance for different positions of an electrode, and different fiber arrays, could be examined. Appendix D contains the computer program listing that was used in this investigation.

4.2.1 Calculation of Jitter for Use in the Model

Many of the single fiber electromyographic recordings from normal muscle show the interval between the action potentials of two muscle fibers at consecutive discharges, the interpotential interval, to be grouped about a stable mean. This interpotential interval has been analysed during recordings in which the mean did not vary and tests show the distribution may be considered as approximately Gaussian [99]. An estimate of the standard deviation (s_D) in the interpotential inter-

val for a finite number of action potential discharges N, (N >> 1) is defined as

$$s_D = \sqrt{\frac{\sum_{i=1}^{N} (D_i - \overline{D})^2}{N}} \tag{4.1}$$

Where D_i is the interpotential interval of the ith discharge, and $\overline{D}$ is the mean interpotential interval. However, Ekstedt [76] reports findings in which the mean time interval between the potentials, would gradually increase or decrease during several hundred discharges. In the presence of slow variations or trends in the mean interpotential interval during measurement, the standard deviation would be overestimated. In a recent paper, Ekstedt et al. [99] present the different methods of expressing the jitter. In conclusion, they suggest that the Mean Consecutive Difference should be used as the measurement of the jitter. This is defined as follows.

$$MCD = \frac{\sum_{i=1}^{N-1} |D_i - D_{i-1}|}{N - 1} \tag{4.2}$$

where D_i is the interpotential interval of the ith discharge. For a Gaussian distribution without trends, an estimate of the standard deviation of the interpotential interval is given by [100]:-

$$s_D = 0.886 \times MCD \tag{4.3}$$

In the normal biceps brachii muscle the mean value of jitter reported by Stålberg et al. [92] from twenty-seven experimental subjects is 15.7 μsec expressed as the Mean Consecutive Difference.

To incorporate the jitter effect in the simulation model of the motor unit, the contribution of one single fiber action potential must be

calculated. Consider two action potentials such that their mean position from an arbitrary axis is a_1 and a_2 as shown in Fig. 4.2. Let x_i and y_i be the intervals from the mean position of each potential due to 'jittering' at the ith discharge. The interpotential interval is given by

$$\begin{aligned} D_i &= (a_2 + y_i) - (a_1 + x_i) \\ &\equiv (a_2 - a_1) + (y_i - x_i) \\ &= K + (y_i - x_i) \end{aligned}$$

An estimate of the mean interpotential interval denoted by $\overline{D}$ is

$$\begin{aligned} \overline{D} &= \frac{1}{N} \sum_{i=1}^{N} (K + y_i - x_i) \\ &= \frac{1}{N} \sum_{i=1}^{N} K + \frac{1}{N} \sum_{i=1}^{N} y_i - \frac{1}{N} \sum_{i=1}^{N} x_i \\ &= K \quad \text{by definition of the mean.} \end{aligned}$$

Since $\sigma_D^2 = E[D^2] - \mu_D^2$ where μ_D and σ_D^2 are the population mean and variance respectively and the estimate of the variance of the interpotential interval is s_D^2 then,

$$\begin{aligned} s_D^2 &= \frac{1}{N} \sum_{i=1}^{N} (D_i)^2 - K^2 \\ &= \frac{1}{N} \sum_{i=1}^{N} (K + (y_i - x_i))^2 - K^2 \\ &= \frac{2}{N} \sum_{i=1}^{N} K(y_i - x_i) + \frac{1}{N} \sum_{i=1}^{N} (y_i - x_i)^2 \end{aligned}$$

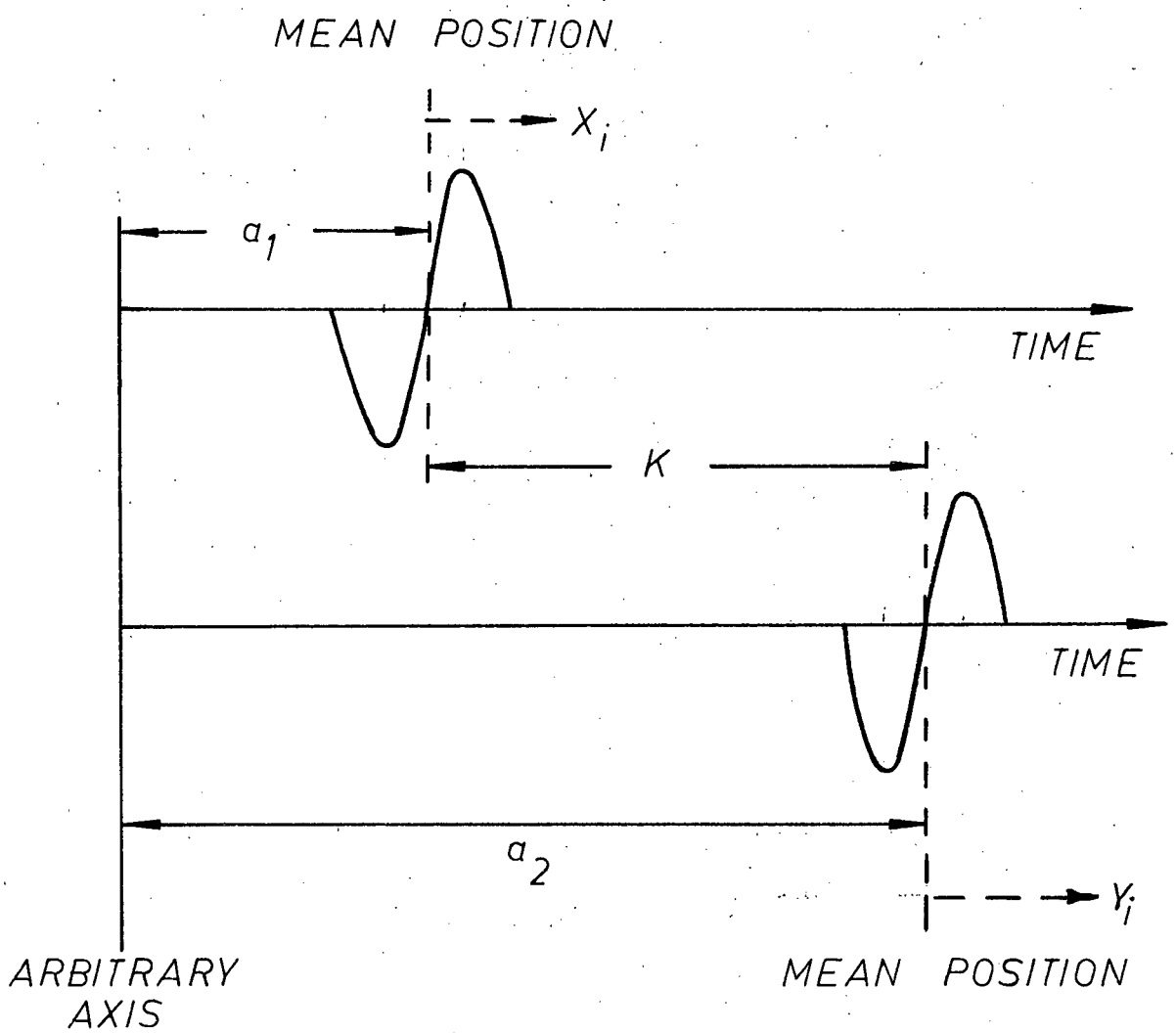


FIG. 4.2 CALCULATION OF ELECTROMYOGRAPHIC JITTER

the first term is zero and the second can be written

$$s_D^{\,2} = \frac{1}{N}\sum_{i=1}^{N} y_i^{\,2} + \frac{1}{N}\sum_{i=1}^{N} x_i^{\,2} + \frac{2}{N}\sum_{i=1}^{N} x_i y_i \,.$$

If we assume that x_i and y_i are independent, i.e. the end-plate delay for one muscle fiber is not dependent on any other, then

$$\frac{1}{N}\sum_{i=1}^{N} x_i y_i = 0$$

The estimate of the variance of x, denoted by $s_x^{\,2}$ is

$$\frac{1}{N}\sum_{i=1}^{N} x_i^{\,2} \qquad \text{(as the mean is zero)}$$

Then, under the assumption that x and y are both identically Gaussian distributed,

$$s_x^{\,2} = \frac{1}{2}\, s_D^{\,2}$$

$$s_x = \frac{s_D}{\sqrt{2}}$$

From equation (4.3), finally for a Gaussian distribution,

$$s_x = \frac{.886}{\sqrt{2}} \times \text{MCD} \qquad (4.4)$$

Using the mean value of MCD in the normal biceps brachii muscle,

$$s_x = 9.8\ \mu\text{sec.}$$

Thus a Gaussian scatter of the single fiber action potentials, with standard deviation of 9.8 μsec was applied for each successive generation of the motor unit potential.

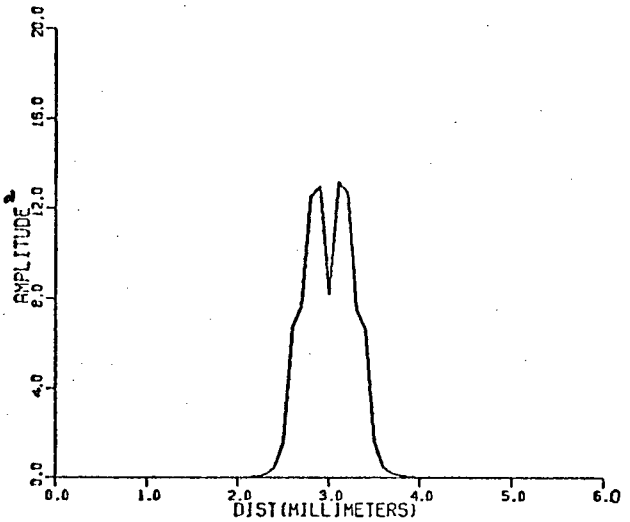
4.2.2 Circular Array of Fibers

In order to examine the effects of jitter on variance detected by fibers at different distances, the variance of the summated potential along the axis of a cylinder of fiber dipoles was computed. Several circular arrays of dipoles, representing muscle fibers, were formed such that the number of dipoles in a circle was proportional to the radius. The muscle fiber diameter was chosen to be 0.80 μm, and each circular array contained the maximum number of fibers for that circle radius. Electromyographic jitter was applied to each single muscle fiber potential as described in section 4.2.1, and the summated action potential was computed along the axis of the array. For each array, an ensemble of 50 summated potentials was generated and the average of the ensemble calculated. The variance was then computed as a function of distance along the muscle fiber axis (related to time by the conduction velocity which was chosen to be 4.1 m/sec), according to the equation

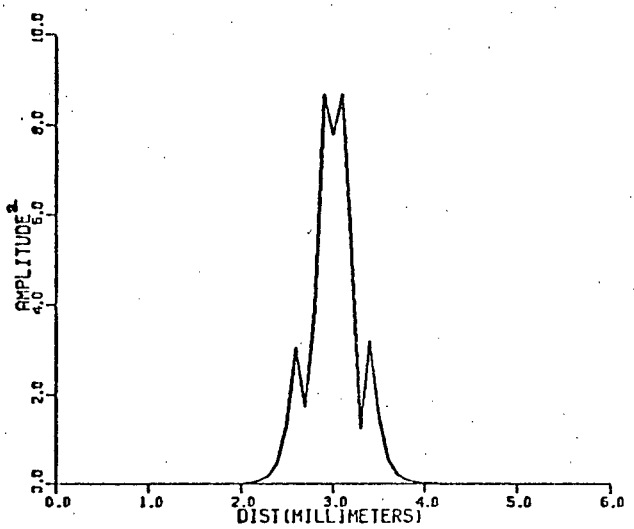
$$S(d)^2 = \frac{1}{M} \sum_{i=1}^{M} (P_i(d)) - \overline{P}(d))^2 \qquad (4.5)$$

where $P_i(d)$ is the amplitude of the potential at the ith generation, $\overline{P}(d)$ is the amplitude of the averaged potential and d represents the discrete distance steps [d, d+0.1, d+0.2 ... in mm] from an arbitrary reference point. M is the number of generated summated potentials in the ensemble. Appendix E contains a listing of the program which computes the averaged potential, the variance of the ensemble and plots the results.

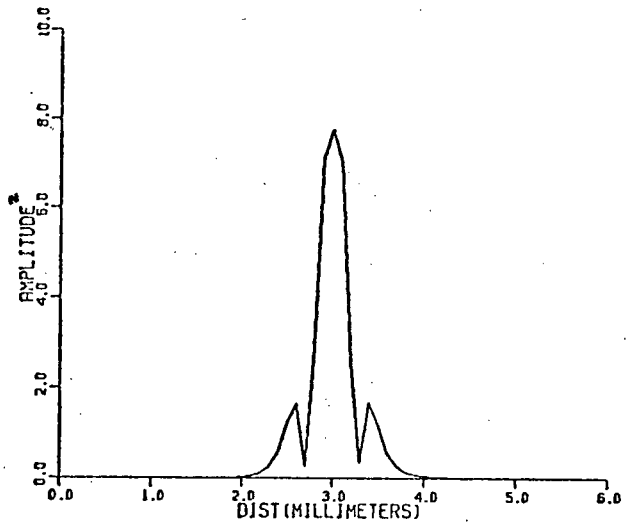
Fig. 4.3 shows the plots of the variance for various radii of the circular array. The contribution of the jitter to the variance of the summated potential, for different fiber distances from the sampling point, is thus observed. The jitter of fiber potentials close to the



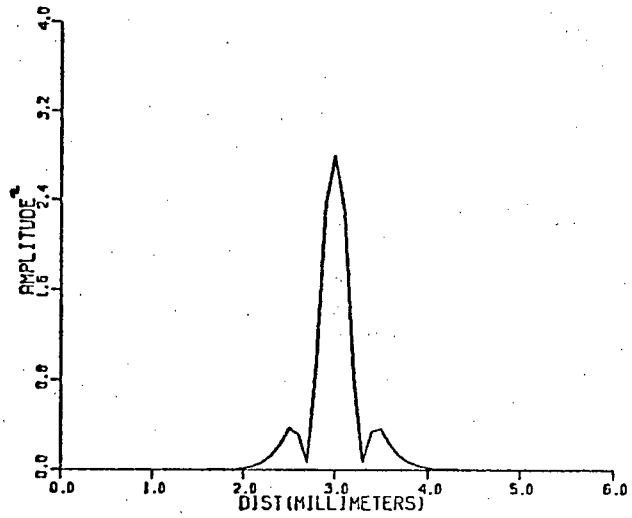
(a)
R = 0.1 mm
F = 8



(b)
R = 0.15 mm
F = 12

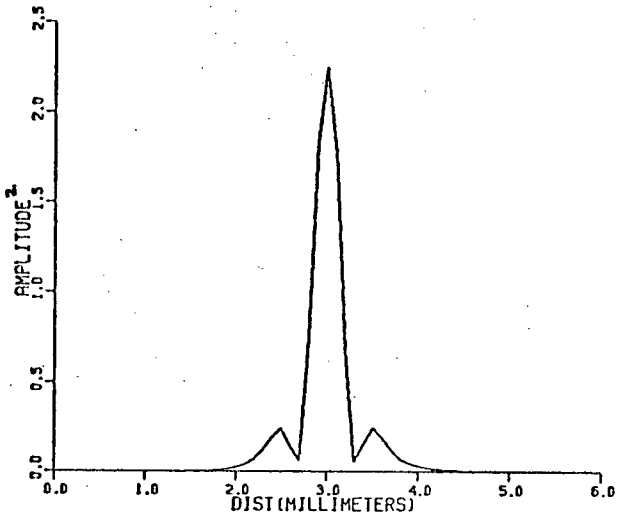


(c)
R = 0.2 mm
F = 16

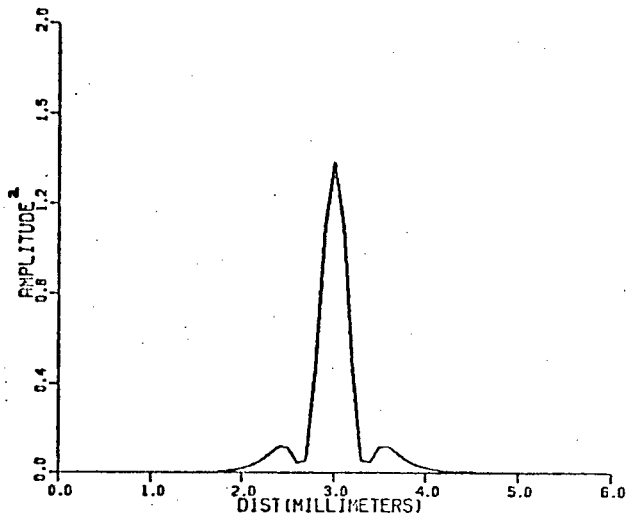


(d)
R = 0.25 mm
F = 20

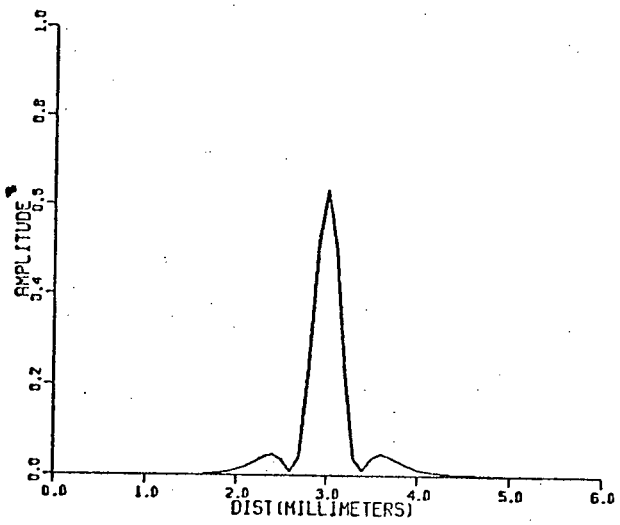
FIG. 4.3 VARIANCE IN AN ENSEMBLE OF SUMMATED POTENTIALS COMPUTED AT THE CENTER OF CIRCLES RADIUS R.
F IS THE NUMBER OF MUSCLE FIBERS IN THE CIRCLE.
ORDINATE: AMPLITUDE OF VARIANCE (IN ARBITRARY UNITS - NOTE CHANGE IN SCALES)
ABSCISSA: DISTANCE ALONG THE MUSCLE FIBER (RELATED TO TIME BY THE CONDUCTION VELOCITY OF THE ACTION POTENTIAL).



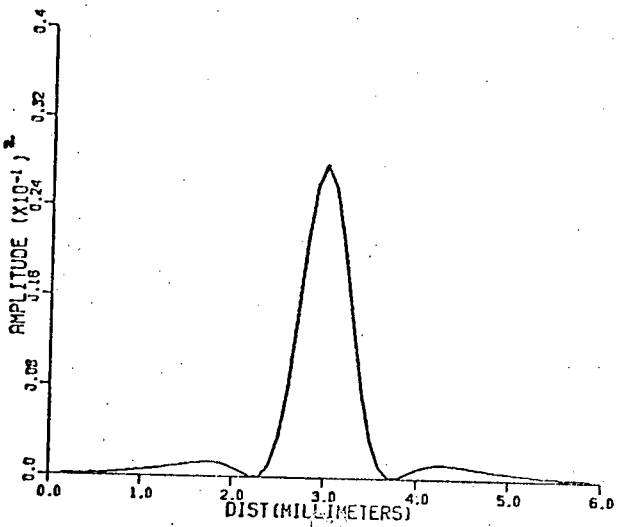
(e)
R = 0.3 mm
F = 24



(f)
R = 0.35 mm
F = 28

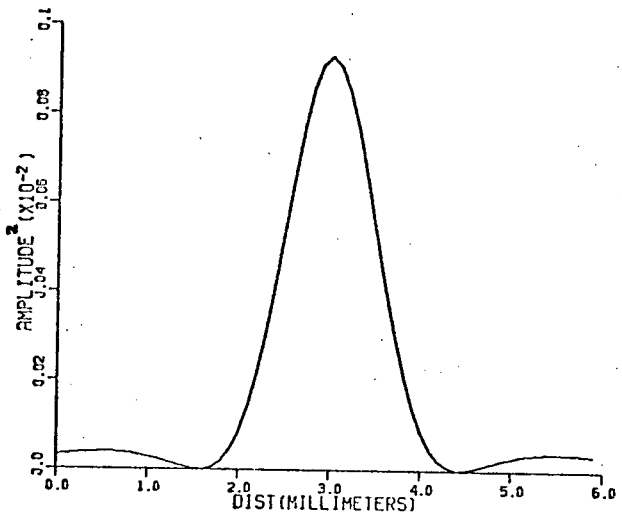


(g)
R = 0.4 mm
F = 32

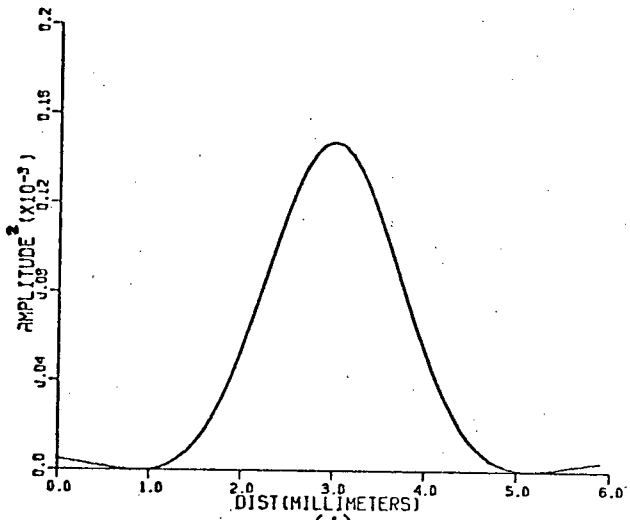


(h)
R = 1.0 mm
F = 78

FIG. 4.3 (Cont'd)



(i)
R = 2.0 mm
F = 156



(j)
R = 3.0 mm
F = 236

FIG. 4.3 (Cont'd)

sample point, cause two peaks in the variance (Fig. 4.3 (a)). As the radius is increased, the two peaks form one peak and side lobes are formed (e.g. Fig. 4.3 (d)), while at still greater distances, the variance becomes very small and has only one distinct peak.

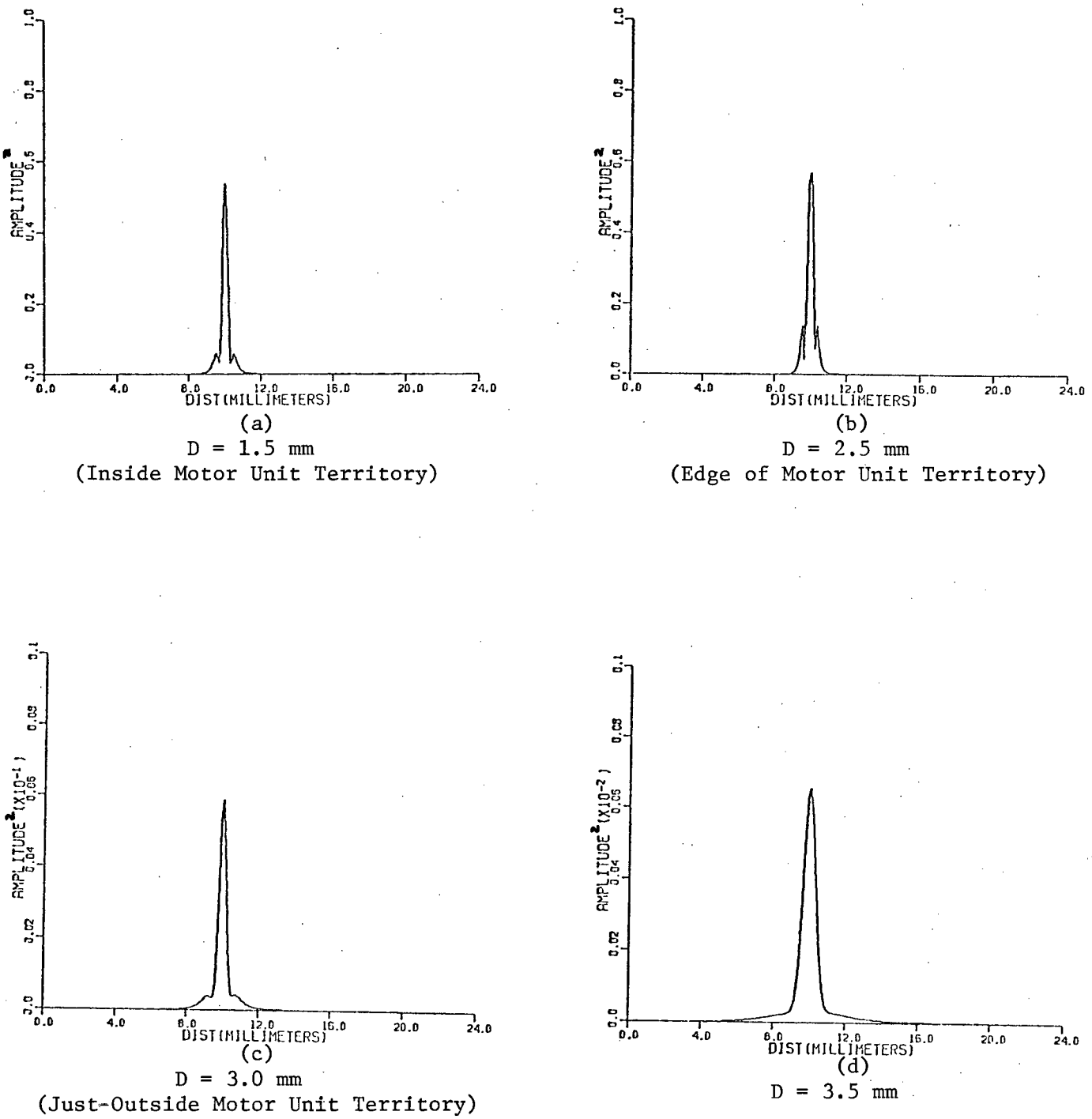
4.2.3 Motor Unit Fiber Array

4.2.3.1 No Axial Potential Dispersion

The variance generated by the anatomical model of the motor unit was next examined. Electromyographic jitter was applied to the single fiber potentials of the random array of 163 fibers described in section 3.2.2, which was used to represent a motor unit in the normal human brachial biceps muscle. Different, Gaussianly distributed (σ = 9.8 μsec) jitter values were applied to the single fiber potentials for each successive generation of the motor unit potential. An ensemble of 50 motor unit potentials was computed for various distances from the motor unit territory. Variance was computed using equation (4.5) and the plotted results are shown in Fig. 4.4. The peak variance as a function of distance outside the motor unit territory has been graphically represented in Fig. 4.5.

4.2.3.2 With Axial Potential Dispersion

Recall from Chapter III that to account for different arrival times at the electrode of the fiber potentials due to conduction time differences in the terminal nerve endings, differences in the axial position of the motor end plates and differences in conduction velocity of the muscle fibers, a Gaussian scatter of the potentials was assumed. Gaussian scatter with a standard deviation of 0.7 mm was applied to the fiber potentials in the motor unit array, together with electromyographic jitter for each successive motor unit potential generated. Fig. 4.6



(a)
D = 1.5 mm
(Inside Motor Unit Territory)

(b)
D = 2.5 mm
(Edge of Motor Unit Territory)

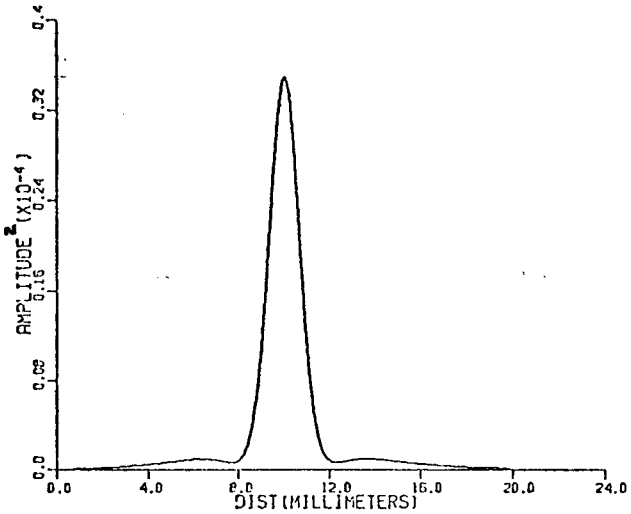
(c)
D = 3.0 mm
(Just-Outside Motor Unit Territory)

(d)
D = 3.5 mm

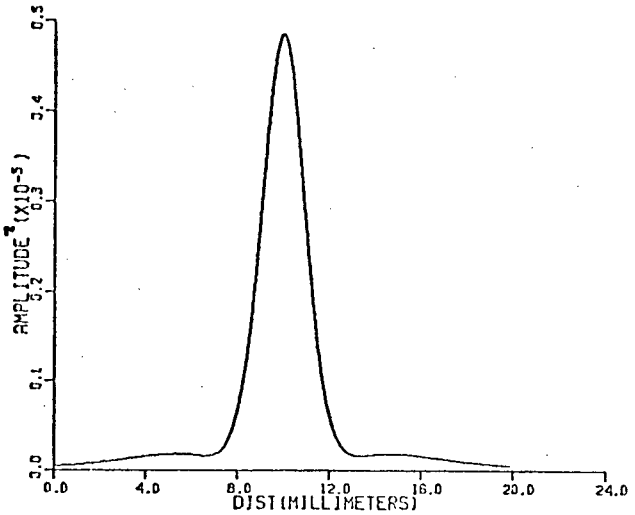
FIG. 4.4 VARIANCE IN THE MOTOR UNIT POTENTIAL AMPLITUDE DUE TO ELECTROMYOGRAPHIC JITTER (D IS DISTANCE FROM THE CENTER OF THE MOTOR UNIT TERRITORY).

ORDINATE: AMPLITUDE OF VARIANCE (IN ARBITRARY UNITS - NOTE SCALE CHANGES)

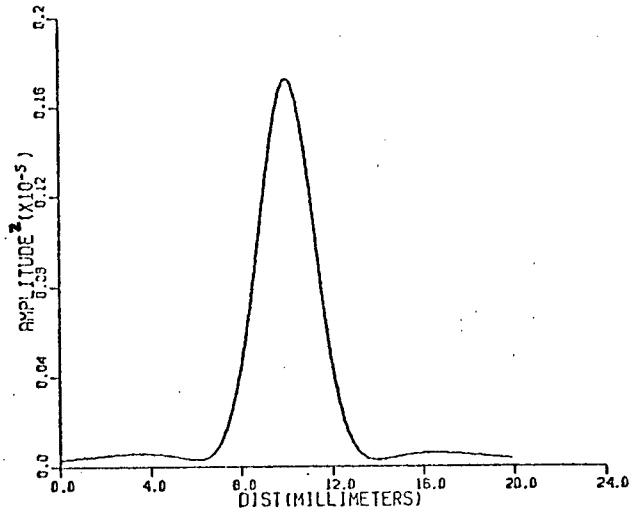
ABSCISSA: DISTANCE ALONG THE MUSCLE FIBER (RELATED TO TIME BY THE CONDUCTION VELOCITY OF THE ACTION POTENTIAL).



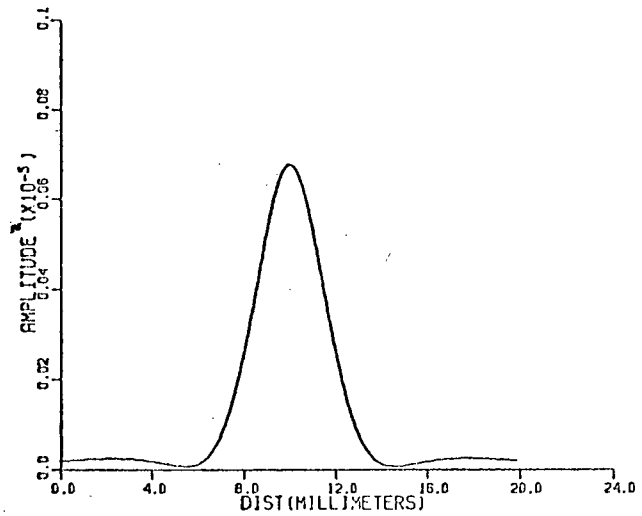
D = 4.5 mm



D = 5.5 mm



D = 6.5 mm



D = 7.5 mm

FIG. 4.4 (Cont'd)

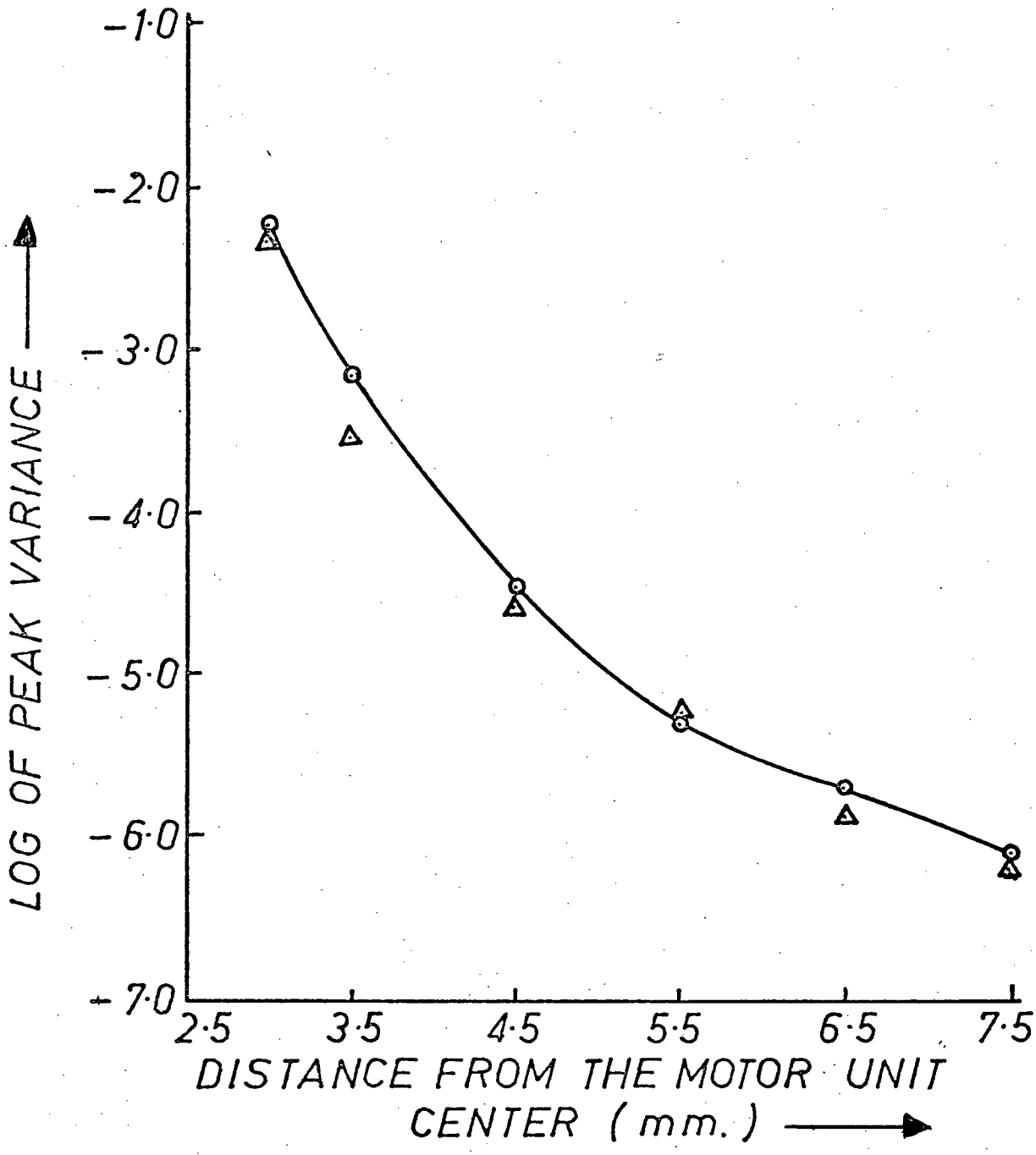


FIG. 4.5 PEAK VARIANCE DUE TO ELECTROMYOGRAPHIC JITTER AS A FUNCTION OF DISTANCE OUTSIDE THE MOTOR UNIT TERRITORY.

⊙ NO AXIAL DISPERSION

△ AXIAL DISPERSION WITH STANDARD DEVIATION = 0.7 mm

shows the variance due to these effects at several distances from the motor unit territory. In Fig. 4.5 the peak variance with potential scatter as a function of distance outside the motor unit area, is compared with no Gaussian axial scatter.

4.3 Experimental Results From Human Subjects

4.3.1 Determination of the Number of Motor Unit Potentials

In order to compare the experimental data with the simulation results, it was desired to determine the number of motor unit potentials with which to estimate the variance. Confidence intervals for the true variance of a normal variate can be found using the chi - square distribution [101]. If there are n motor unit potentials in an ensemble, the confidence interval for the true variance σ_A^2 of the amplitude at each sample point in time is

$$\frac{nS_A^2}{\chi^2_{P_1}(n-1)} \leq \sigma_A^2 \leq \frac{nS_A^2}{\chi^2_{P_2}(n-1)} \tag{4.6}$$

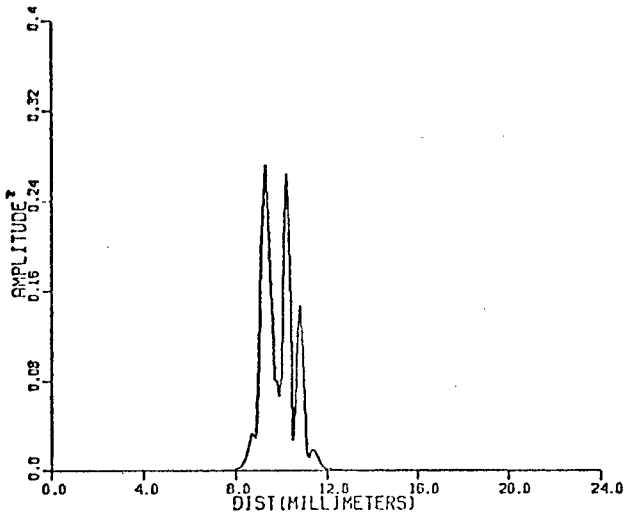
where S_A^2 is the variance of the ensemble amplitude for each sample point in time and $\chi^2_{P_1}(n-1)$ is the value of chi - squared for (n-1) degrees of freedom evaluated at $P = P_1$ where

$$100\%\ P_1 = 1/2\ (100\% - \text{confidence level})$$

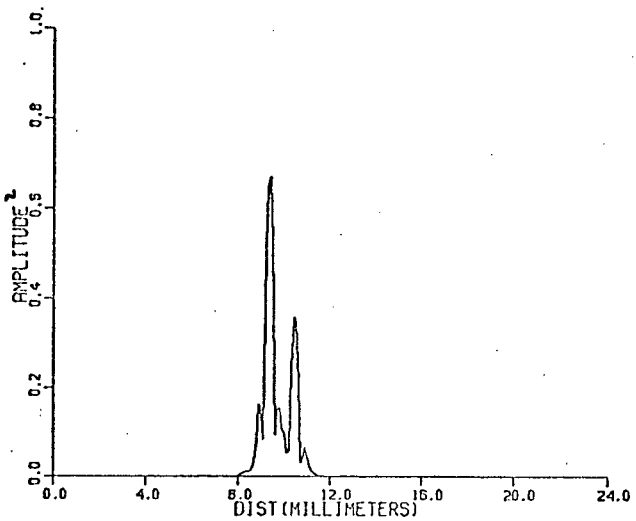
Similarly, $\chi^2_{P_2}(n-1)$ is evaluated for (n-1) degrees of freedom for $P = P_2$ where

$$100\%\ P_2 = 1/2\ (100\% + \text{confidence level}).$$

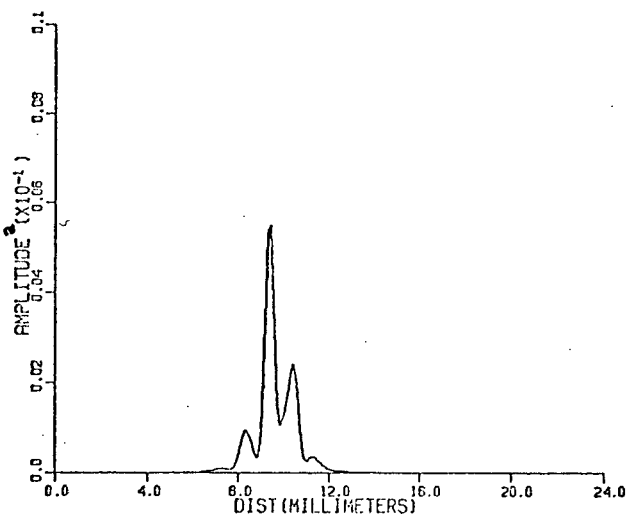
Using the interactive graphics routine described in section 2.4.1 single motor unit potentials were extracted from the digitized



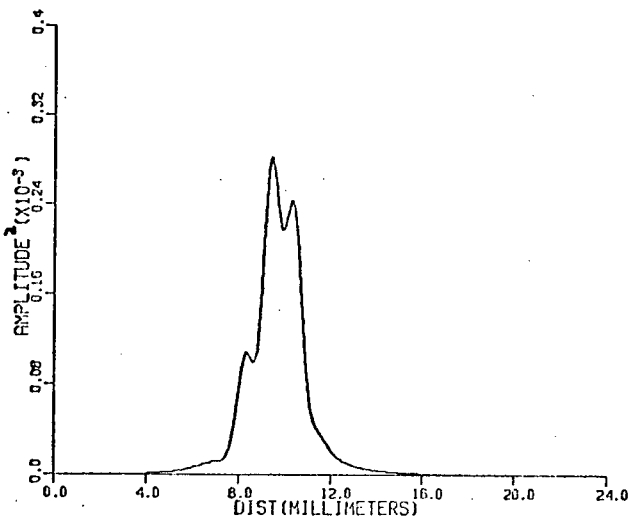
(a)
D = 1.5 mm
(Inside Motor Unit Territory)



(b)
D = 2.5 mm
(Edge of Motor Unit Territory)



(c)
D = 3.0 mm
(Just Outside Motor Unit Territory)

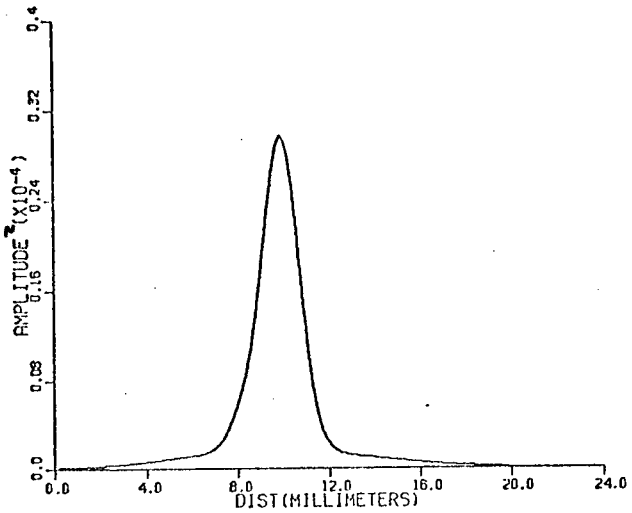


(d)
D = 3.5 mm

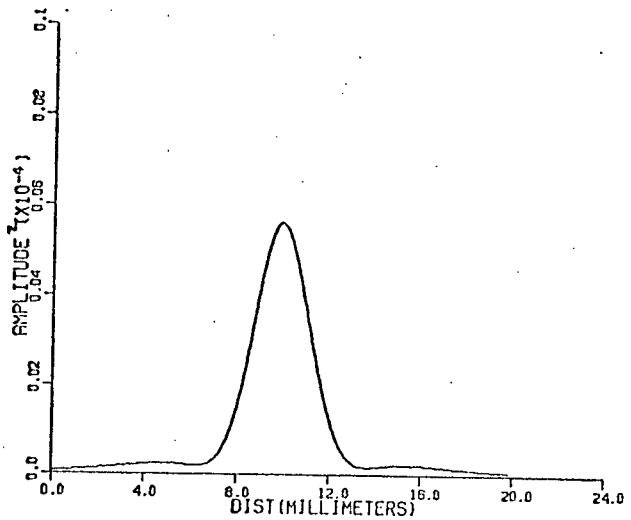
FIG. 4.6 VARIANCE IN THE MOTOR UNIT POTENTIAL DUE TO EFFECTS OF (1) ELECTROMYOGRAPHIC JITTER AT EACH SUCCESSIVE GENERATION (2) SCATTER OF DIPOLES ($\sigma\sigma$= 0.7 mm).

ORDINATE: AMPLITUDE OF VARIANCE (IN ARBITRARY UNITS - NOTE SCALE CHANGES).

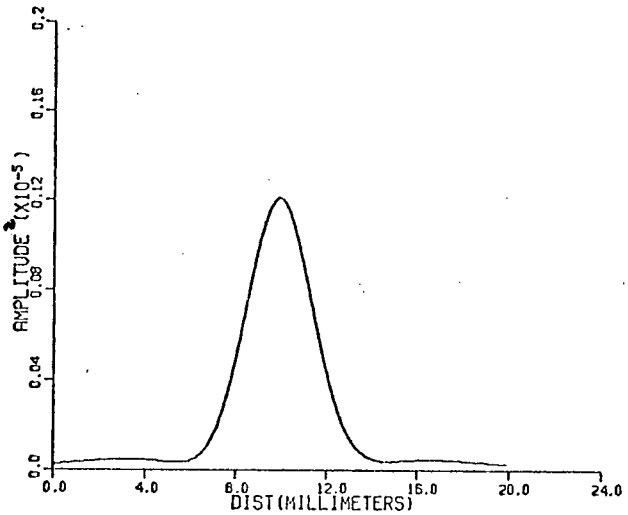
ABSCISSA: DISTANCE ALONG THE MUSCLE FIBER (RELATED TO TIME BY CONDUCTION VELOCITY OF THE ACTION POTENTIAL).



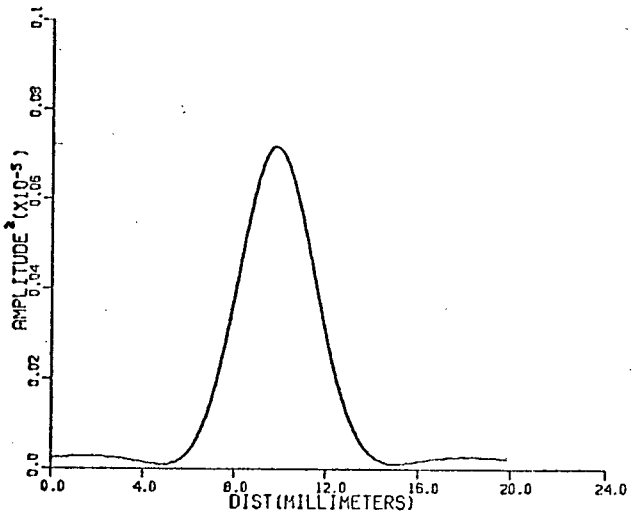
(e)
D = 4.5 mm



(f)
D = 5.5 mm



(g)
D = 6.5 mm



(h)
D = 7.5 mm

FIG. 4.6 (Cont'd)

time series EMG waveform recorded from normal subjects, and the averaged motor unit potential calculated by the procedure outlined in Fig. 2.3 (c). The variance for each amplitude sample was then computed. Under the assumption that each amplitude sample of the motor unit potential may be considered as a normal variate, the confidence interval for a 95% confidence level was found using the chi - squared distribution. To determine the number of motor unit potentials to use for variance estimation, the largest confidence interval (worst case condition) was selected and plotted as a function of the number of potentials. The results of this investigation from three normal subjects are given in Fig. 4.7 to Fig. 4.9. The results clearly indicate the desirability of using greater than fifty motor unit potentials to estimate the variance in an ensemble. Eighty motor unit potentials were chosen.

It should be pointed out, in the implementation of a clinical system for variance estimation it is desirable that the number of motor unit potentials required for analysis should be small for practical reasons. Routine examinations would be less demanding for the patient and the clinician may carry out the test in a short period of time.

4.3.2 Normal Subjects

Using the interactive graphics routine, eighty potentials from the same single motor unit were extracted from the EMG time series waveform that had been recorded from the biceps brachii muscle of a normal subject. After alignment, the averaged motor unit potential was computed and the variance was calculated for comparison with the predictions from the simulation model. During the recording of the EMG activity no attempt was made to examine only diphasic potentials such as given by the simulation model (see section 4.4). The plots of the

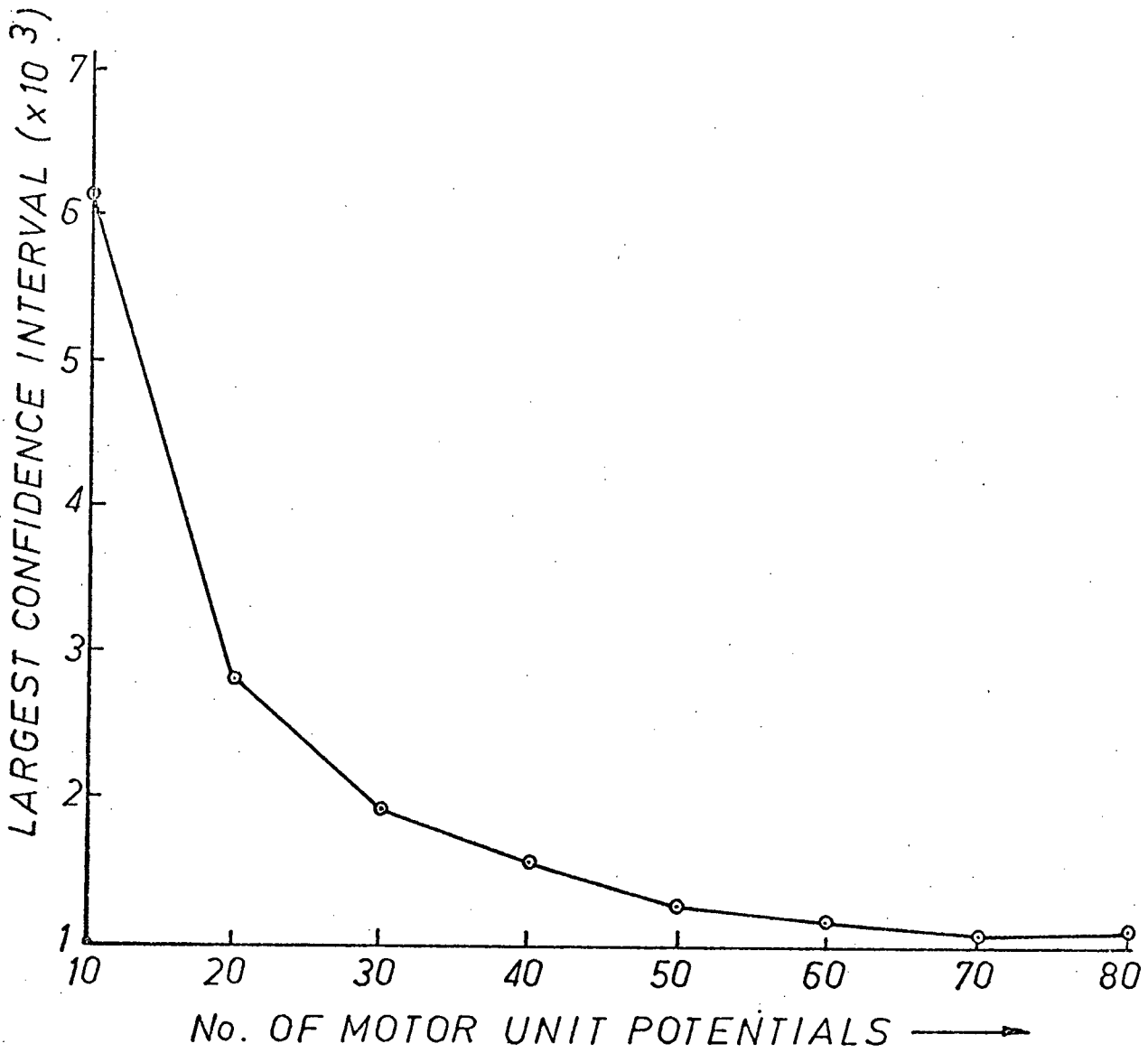


FIG. 4.7 EFFECT OF INCREASING THE NUMBER OF MOTOR UNIT POTENTIALS TO ESTIMATE VARIANCE, ON THE CONFIDENCE INTERVAL FOR A 95% CONFIDENCE LEVEL. SUBJECT T.M.

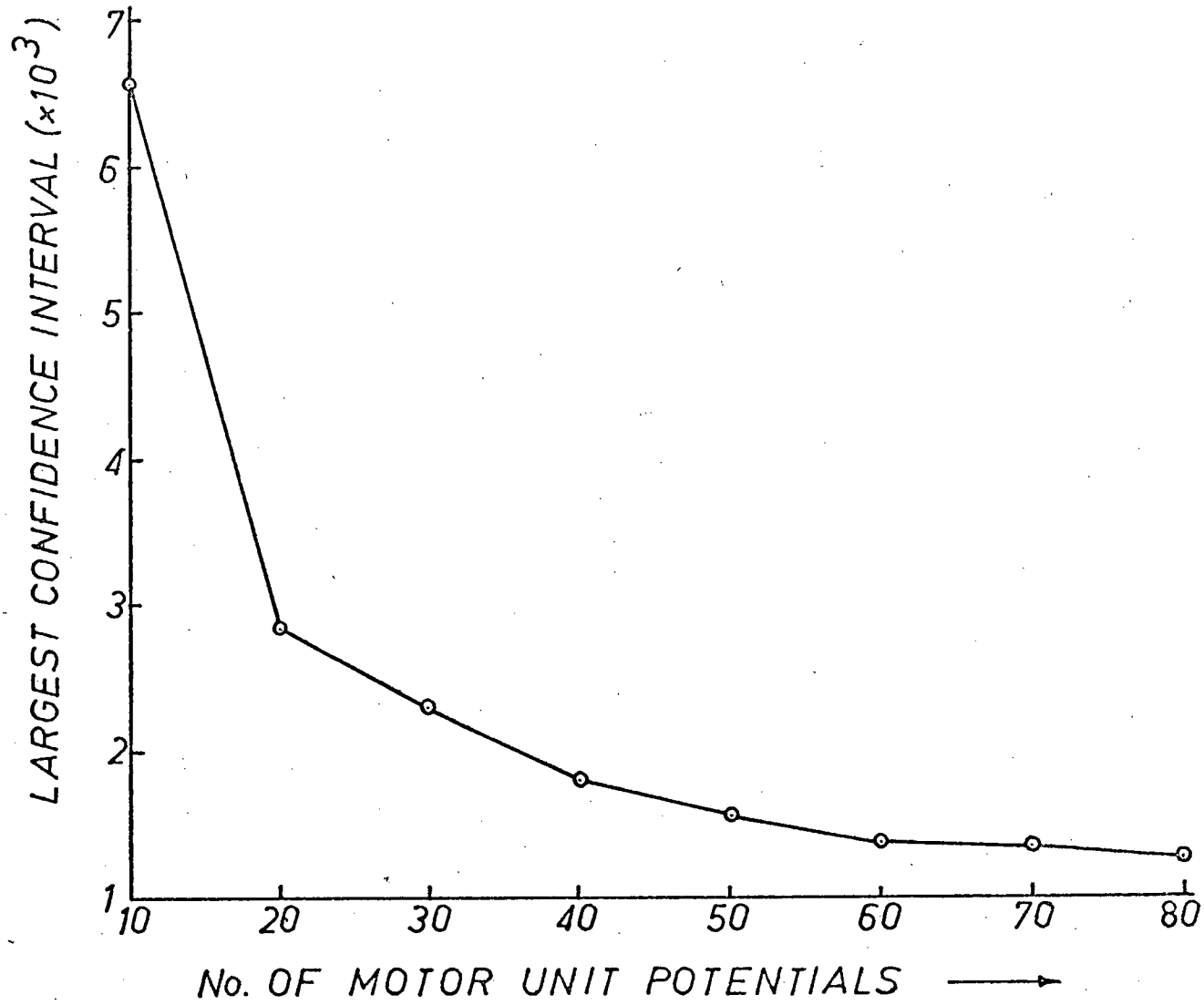


FIG. 4.8 AS IN FIG. 4.7. SUBJECT P.B.

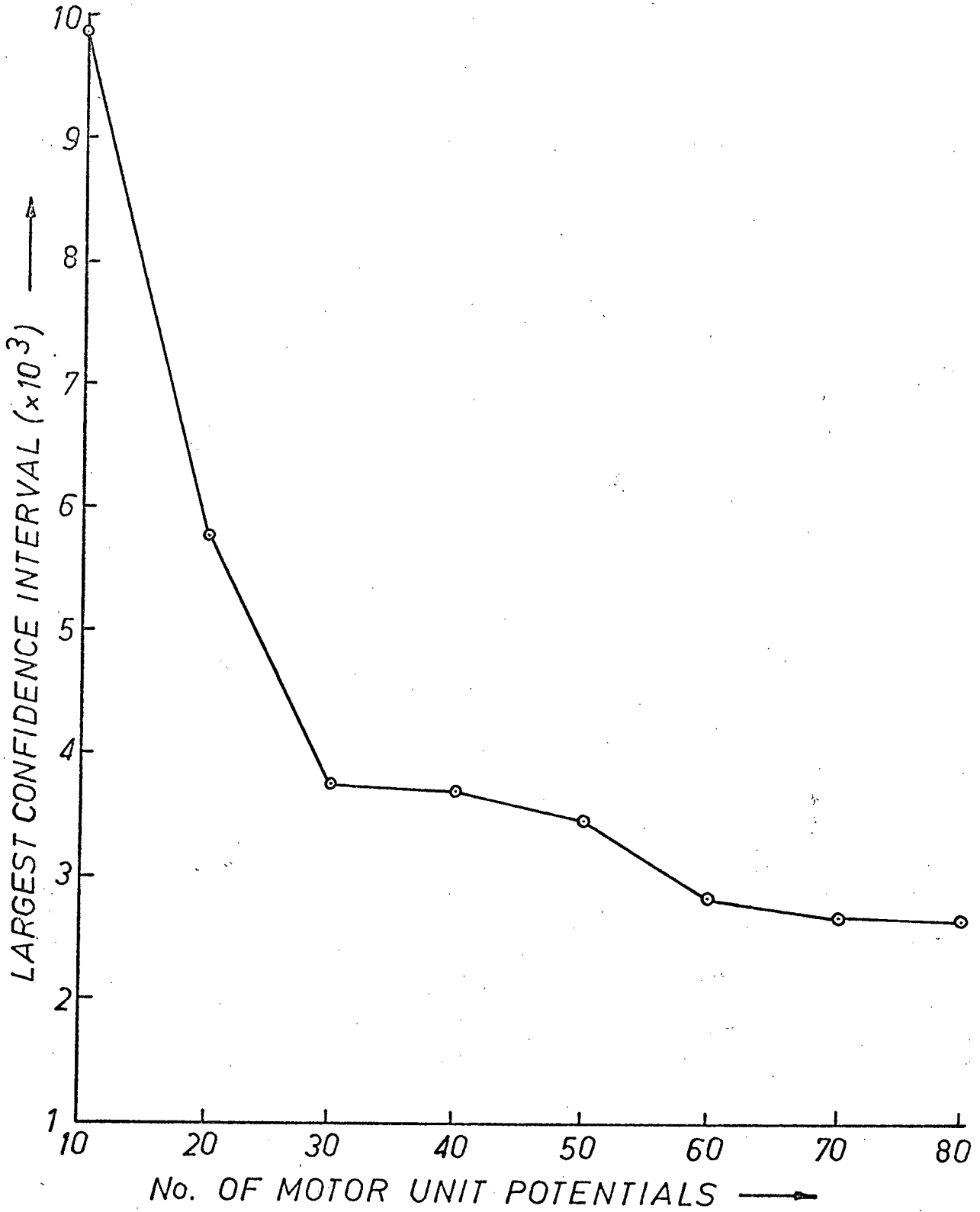


FIG. 4.9 AS IN FIG. 4.7. SUBJECT P.L.

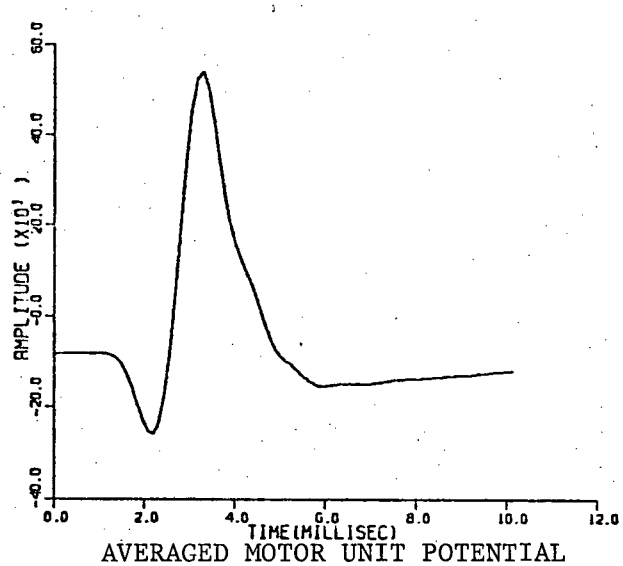
variance against time, together with the averaged motor unit potential are shown for 4 different motor units (3 subjects) in Fig. 4.10 to Fig. 4.13.

4.3.3 Patients with Muscle Disease

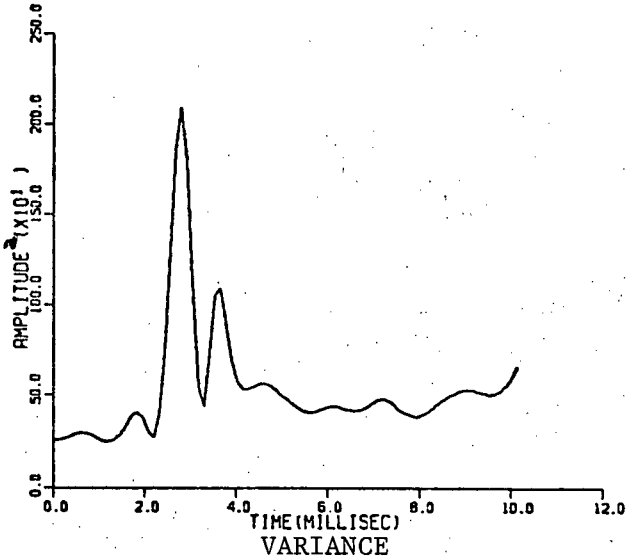
In patients with myasthenia gravis, Ekstedt and Stålberg [94] and Stålberg et al. [98] found that the jitter in many single fiber potential pairs was highly increased from the normal. This jitter was usually above 100 μsec (MCD). Using equation (4.4), an increased jitter of 100 μsec (MCD) was incorporated into the simulation model and the variance of the motor unit potential calculated for various points inside and outside the motor unit area. The results reveal that there is a large increase in peak variance within and near the motor unit territory. Fig. 4.14 shows a typical averaged motor unit potential computed at a point inside the motor unit territory and also the amplitude variance calculated from the motor unit potential ensemble (units of amplitude as in the normal model case).

Repeating the procedure of the preceeding section for a patient with myasthenia gravis, the experimental results follow the predictions of the model. Fig. 4.15 shows an averaged motor unit potential which is diphasic and appears to be clinically normal yet the peak variance is much greater than that from the normal subjects (same units of amplitude as recorded in the normal experimental subjects).

Initial results from the polymyositis EMG data show there are several large peaks of variance. Further investigations are required to quantify this effect.



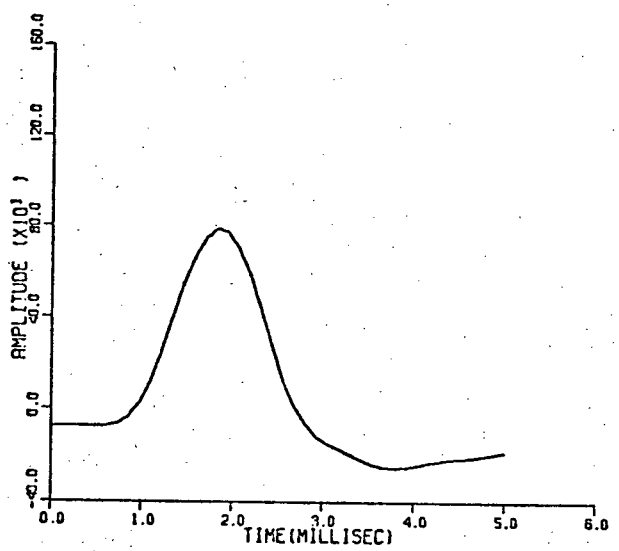
AVERAGED MOTOR UNIT POTENTIAL



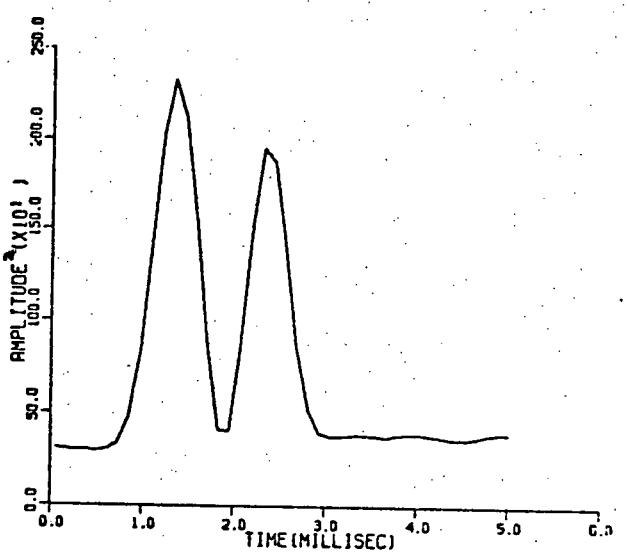
VARIANCE

FIG. 4.10 MOTOR UNIT POTENTIAL VARIANCE IN AMPLITUDE FROM NORMAL SUBJECT T.M. (RECORDINGS FROM THE BICEPS BRACHII MUSCLE)

$$\text{Amp(mV)} = 2.44 \times 10^{-4} \times \text{Amp(units)}$$

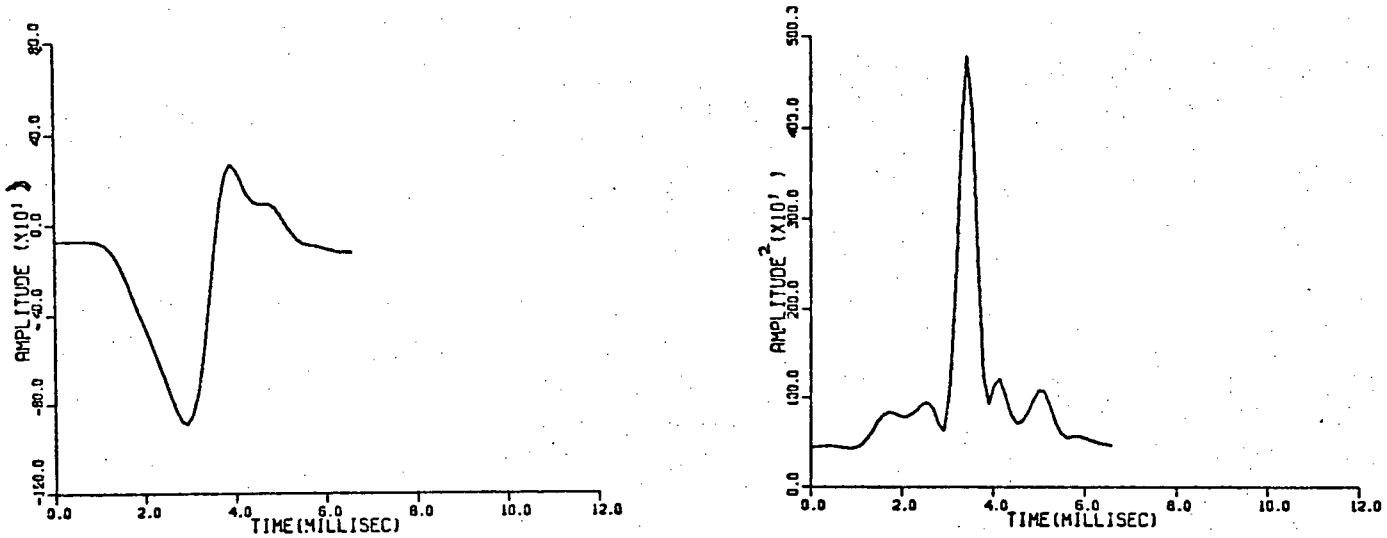


AVERAGED MOTOR UNIT POTENTIAL



VARIANCE

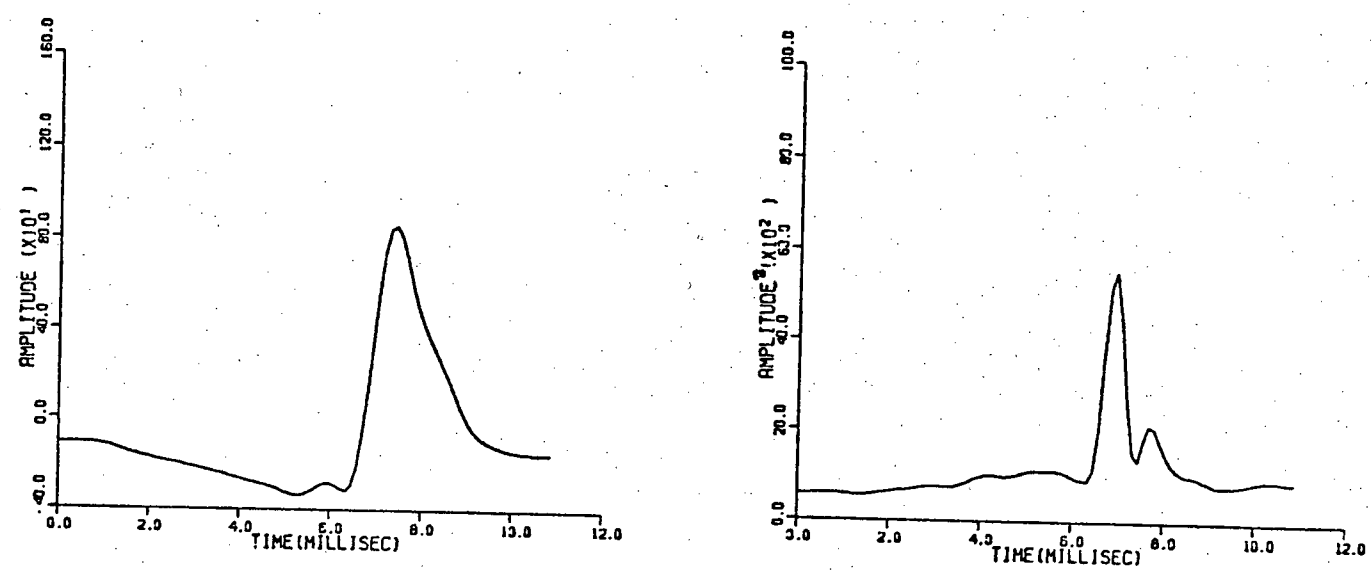
FIG. 4.11 AS IN FIG. 4.10. SUBJECT P.B.



AVERAGED MOTOR UNIT POTENTIAL VARIANCE

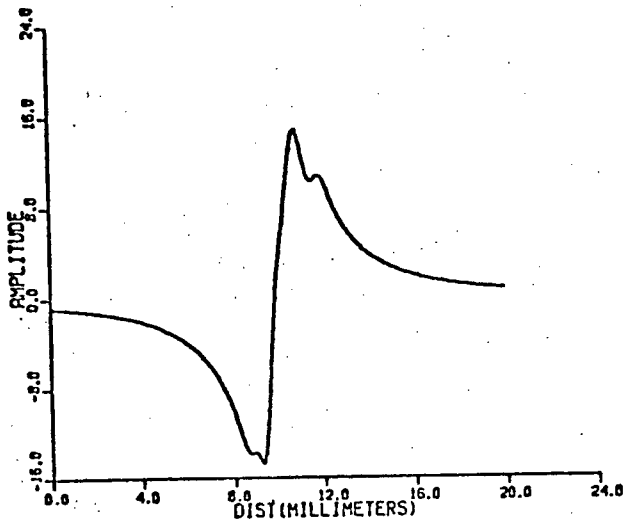
FIG. 4.12 MOTOR UNIT POTENTIAL VARIANCE IN AMPLITUDE FROM NORMAL SUBJECT P.L. (RECORDING FROM THE BICEPS BRACHII MUSCLE)

$$\overline{\text{Amp}}(\text{mV}) = 2.44 \times 10^{-4} \times \text{Amp(units)}$$

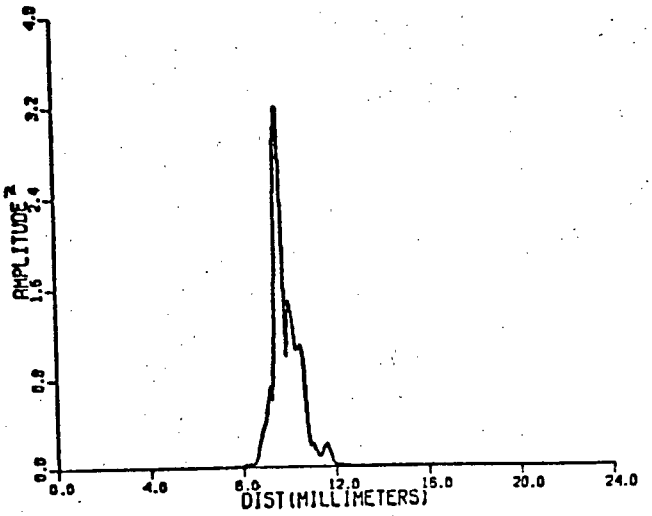


AVERAGED MOTOR UNIT POTENTIAL VARIANCE

FIG. 4.13 AS IN FIG. 4.12. SUBJECT P.L.

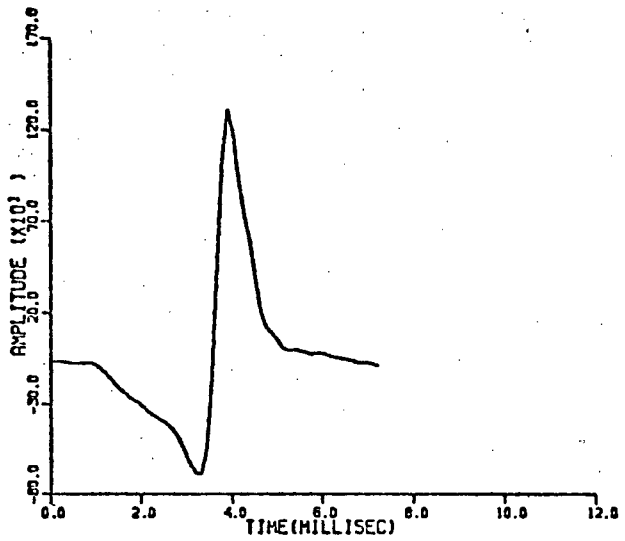


(a) AVERAGE MOTOR UNIT POTENTIAL

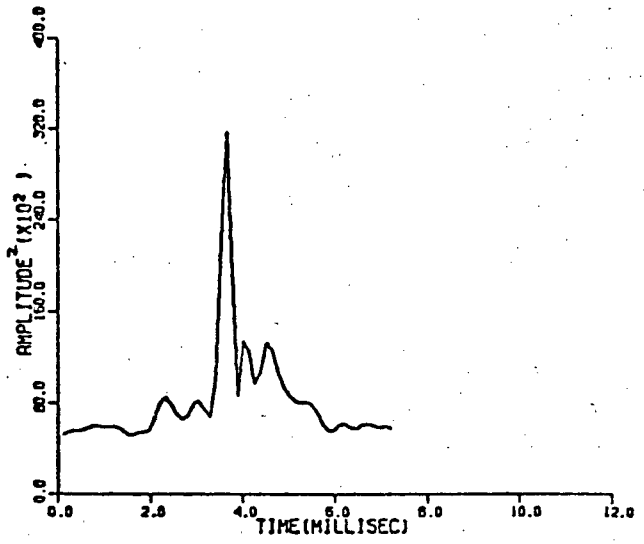


(b) VARIANCE

FIG. 4.14 MODELLING OF MYASTHENIA GRAVIS - THE VARIANCE DUE TO INCREASED JITTER.
AXIAL DIPOLE DISPERSION = 0.7 mm
JITTER = 100 μsec (MCD)
COMPUTED INSIDE THE MOTOR UNIT AREA



(a) AVERAGED MOTOR UNIT POTENTIAL



(b) VARIANCE

FIG. 4.15 VARIANCE OF MOTOR UNIT AMPLITUDE IN A PATIENT WITH MYASTHENIA GRAVIS

$$\text{Amp(mV)} = 2.44 \times 10^{-4} \times \text{Amp(units)}$$

4.4 Discussion

4.4.1 General Discussion

The computer simulation model of the motor unit potential has been employed to investigate the observation that in each successive firing of the motor unit potential, there is a variation in its shape and that this variability may be increased in certain diseases. Using the techniques of single fiber electromyography other investigators have shown that there is a jitter of the time interval between two fiber action potentials at consecutive discharges. This jitter may be due to some variability in the synaptic delay of the normal end-plate [94]. It is a reasonable assumption that in the normal human muscle the jitter phenomenon would be a factor likely to alter the contour of a summated potential. As an initial approach to the investigation of shape variability of the motor unit potential, this hypothesis was tested by simulating electromyographic jitter and investigating the effect on the variance of the motor unit potentials generated.

It has been reported that the interpotential interval between two fiber potentials may be considered to be approximately Gaussianly distributed [99]. In the literature some interpotential intervals have been found with distributions which had skewness, positive kurtosis or more commonly, negative kurtosis. Some bimodal distributions were also found [102]. The calculation of jitter for use in the model was based on the assumption that the interpotential interval has a Gaussian distribution. The value of jitter used in the model was 15.7 μsec (MCD) which is the mean value reported [92] for the normal biceps brachii muscle from 27 experimental subjects. The range given is 2.6 - 37.1 μsec (MCD).

The effect of jitter on the variance of the summated potentials from a circular array of dipoles was first studied. Using this array structure, the variance caused by jitter of the volume conducted single fiber potentials at different distances was examined. The results indicate that jittering of potentials from a nearby electrode gives rise to several peaks in variance (Fig. 4.3 (b)). At greater distances one variance peak becomes more dominant (Fig. 4.3 (i)). The random fiber array representing a motor unit in human biceps brachii muscle shows that there is a peak variance in the rising edge of the diphasic motor unit potential. This peak variance decreases rapidly with distance outside the motor unit territory (Fig. 4.5). With the introduction of axial dispersion in the manner described in section 4.2.3, some asymmetry in the lobes of the variance plots was observed (Fig. 4.6 (c)).

Before comparing the experimental data with the model, a test was performed on the variance confidence intervals to find the number of motor unit potentials required to estimate variance. The test revealed that for a 95% confidence level at least fifty potentials were sufficient for the estimation. It is desirable for practical reasons to keep this number to a minimum. Ensembles of eighty motor unit potentials were chosen to compute the variance in amplitude for several normal subjects. The variance results clearly agree with model predictions. A large peak of the variance is dominant on the rising edge of the motor unit potential with asymmetrical lobes.

In the modelling of myasthenia gravis, the jitter value was increased but a Gaussian distribution was still assumed. It is more common in the pathological state for the interpotential interval to have a distribution which in not Gaussian [99]. The model however, predicts

a large increase in peak variance. Motor unit potentials recorded from a myasthenic muscle, did show a large increase in peak variance from that of the normal. It must be pointed out that in myasthenia gravis there is also a neuromuscular blocking when the single fiber potential does not fire for several motor unit discharges. This will also add to the variance.

The experimental results show that the baseline of variance is non-zero. This can be modelled in the following way (see Fig. 4.16). Let G represent a deterministic action potential generator. N represents the generator noise due to electromyographic jitter, while B is the biological noise source added by distant motor unit potentials, the recording apparatus and digitizing process. The total variance then will be the sum of the generator noise variance and the biological noise variance which will contribute to the baseline. In the pathological state, the generator noise increases due to increased electromyographic jitter, blocking of single fiber potentials or amplitude variation in successive figures of the single fiber potentials.

4.4.2 Extensions of this Work

The initial results of this investigation should provide exciting stimulus for further work. It has been beyond the scope of this thesis to quantify this variance. A further expansion of the data base is necessary to establish limits for normal variance in motor unit potentials and variance due to pathological conditions. A study of other physiological factors in the generation of motor unit potentials which affect the variance should be undertaken. The model developed in this thesis may be usefully employed for this purpose.

In an attempt to compare normal motor unit potentials recorded

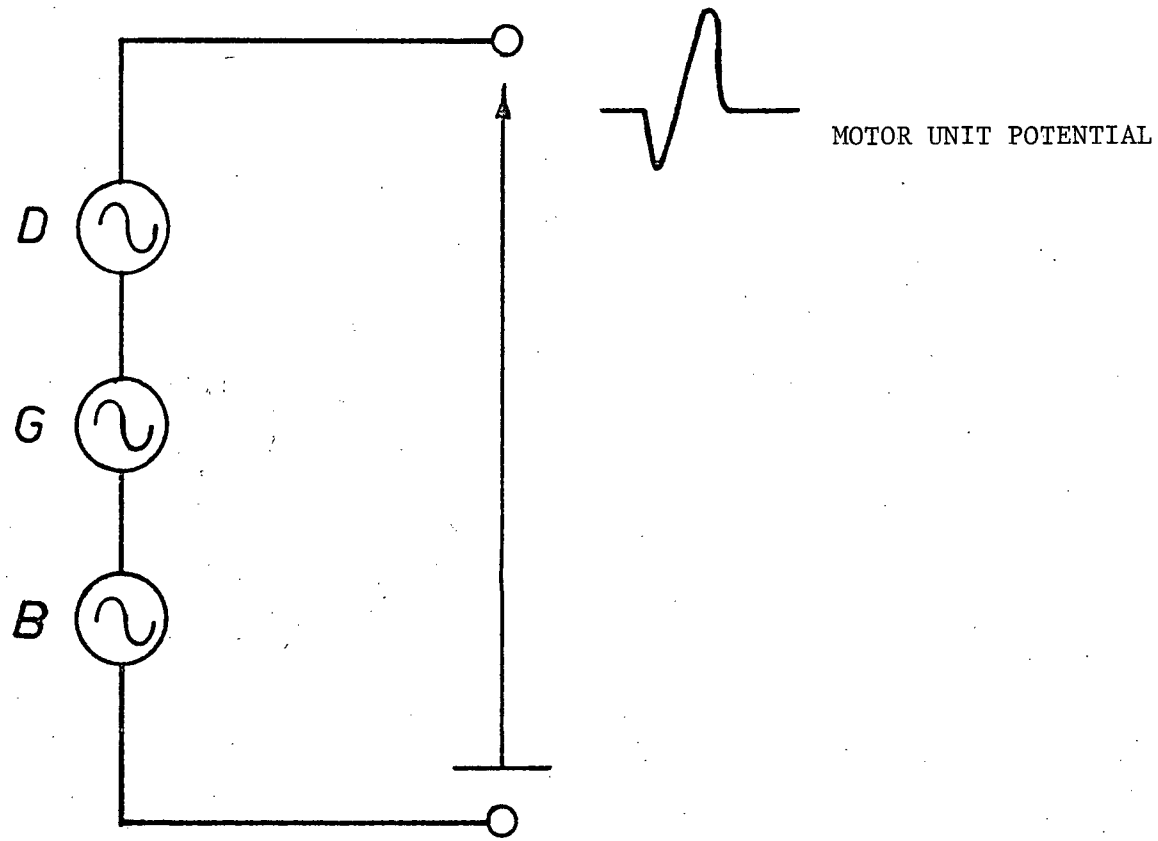


FIG. 4.16 SIMPLE GENERATOR MODEL REPRESENTING THE RECORDED MOTOR UNIT POTENTIAL

from the experimental subject, it was found that many did not have a symmetrical biphasic shape. Some potentials demonstrated a large initial phase (Fig. 4.12) or a large second phase (Fig. 4.10). Several motor unit potentials had an almost monophasic time course (Fig. 4.11, Fig. 4.13). It is fundamental to any further investigation of variance to understand how these shapes of the action potentials are produced. One possible area deserving further exploration is the transformation induced on the motor unit potential by the sampling electrode, and how this electrode distorts the volume conducted electric field.

CHAPTER V

CONCLUSIONS AND DIRECTIONS FOR FURTHER RESEARCH

5.1 Synopsis of the Thesis

The objective of the work described in this thesis was to study the production of the motor unit potential in order that the knowledge gained may be used in the development of practical system for automatic analysis of EMG to aid the clinician in his diagnosis. Methods for the acquisition and subsequent preprocessing of single motor unit potentials were devised. An EMG data base consisting of normal and pathological EMG activity from the biceps brachii muscle has been established and an interactive graphics routine developed to visually extract potentials from the same motor unit for further analysis. A computer model was proposed for the generation of motor unit potentials observed in the clinical EMG examination of the normal biceps brachii muscle. Variations in the peak to peak amplitude and peak to peak duration at different axial potential dispersions were investigated for points at increasing distance from the motor unit axis. This model has also been compared with existing experimental data from multielectrode studies of the muscle. The model was further employed in the investigation of the variation in the shape of the motor unit potential due to the effects of EMG jitter. The acquired experimental data has been analysed and compared with the predictions of the model concerning the variation in shape of normal motor unit potentials due to electromyographic jitter. This investigation initially has demonstrated that in myasthenia gravis, a disease in which electromyographic jitter is increased, the peak variance also increased. Further work has been indicated in this area.

5.2 Contributions of the Research

The author considers the main contributions of this research may be summarized as follows:

(1) A comprehensive review of the methods of automatic analysis in clinical electromyography has been added to the current literature [62] including the proposal of a set of design requirements for automated analysis of EMG signals.

(2) Methods for the acquisition and preprocessing of clinical EMG activity have been established at U.B.C.

(3) An interactive graphics routine to extract single motor unit potentials has been developed. To the knowledge of the author, such a method has not been reported before.

(4) A model for the generation of the motor unit potential based on physiological findings has been proposed and investigated for the first time.

(5) The first investigation into the variation in motor unit potential shape at successive discharges due to electromyographic jitter has been made.

(6) The observation has been made that the variance in motor unit potential amplitude was greatly increased in a case of myasthenia gravis. The reason, it was suggested, was due partially to an increase in electromyographic jitter.

A review of the methods of automatic analysis in clinical EMG revealed that many were unreliable and, with the exception of Willison's, lacked clinical significance. Many of the methods described use heuristically derived indices not based on a fundamental understanding of how disease affects the recorded activity. This research represents a new

approach. A knowledge of the production of the motor unit potential, together with how the disordered physiology affects the fiber potentials, can give more foundation to the selection of the appropriate signal processing techniques.

5.3 Directions for Further Research

The following areas are suggested for further work.

(a) Expansion of the EMG Data Base

(b) Improvements in the method to extract single motor unit potentials.

(c) Further extensions of the motor unit potential model and human experimental verification.

(d) Study of the variance in motor unit potentials in myopathy and neuropathy and of its diagnostic usefulness.

5.3.1 Expansion of the Data Base

The success of any further work in this research is dependent upon the expansion of the EMG data base. One improvement that would be of benefit to this goal is the development of on-line digitization of the EMG data at Vancouver General Hospital. IBM compatible tapes may be taken directly to the U.B.C. computer center for further analysis and the analog stage eliminated.

5.3.2 Interactive Graphics Routine

The extraction of eighty motor unit potentials from the EMG waveform is time consuming and laborious. Thus, computer routines to automatically align the motor unit potentials would leave the operator after initialization with an accept/reject decision.

Application of pattern recognition techniques is necessary for

the extraction of the motor unit potentials in a practical clinical system. In this regard, the computer programs should use a minimum amount of memory such that the system may be mini or microcomputer based.

5.3.3 Motor Unit Potential Model

This model has proved to be useful in the study of the electrical activity from the normal motor unit. The properties of the electrode and subsequent amplifiers must be examined in order that the distortion on the motor unit potential be compensated. Further experimental verification of the model from human subjects, such as amplitude versus duration plots would be beneficial.

5.3.4 Study of Motor Unit Potential Variance

Using the model, pathological processes can be simulated and the effects on variance tested for diagnostic significance.

APPENDIX A

TERMS AND DEFINITIONS OF ELECTROMYOGRAPHY

Some of the more important terms and definitions of electromyography used in this thesis are given below for reference. Further information on the terminology may be obtained from [103] or from any of the more general texts [5,6,7].

A.1 PHYSIOLOGY

A.1.1 Biceps Brachii One of the muscles of the upper arm. It flexes the forearm and it turns the hand so that the palm can face upwards.

A.1.2 Motor end-plate The flat expansion ending a motor nerve fiber where it connects with a muscle fiber and includes portions of nerve and muscle.

A.1.3 Motor Unit This is the functional unit of the neuromuscular system consisting of the anterior horn cell in the spinal cord, its axon and all the muscle fibers innervated by that axon.

A.2 ACTION POTENTIALS

A.2.1 Motor Unit Potential The action potential expressing the activity of that part of a single motor unit which is within the recording range of an electrode.

A.2.2 Total duration As defined by Buchthal [10], this is the time interval between the initial deflection from the baseline and the point at which the terminal deflection again returns to the baseline.

A.2.3 Monophasic action potential An action potential with a deflection to one side of the baseline.

A.2.4 Biphasic action potential An action potential with a deflection first to one side then to the other side of the baseline

(usually a positive-negative sequence).

A.2.5 Polyphasic Action Potential An action potential having more than four phases.

A.3 RECORDING EQUIPMENT

A.3.1 Concentric needle electrode Variations in voltage are measured between the bare tip of an insulated wire, usually stainless steel, or platinum, and the bare shaft of a steel cannula in which it is inserted. The bare tip of the central wire (exploring electrode) is flush with the level of the cannula (reference electrode).

A.3.2 DISA electromyograph Instrument used to amplify and display the EMG signals. It also includes an audio amplifier and speaker to allow acoustic monitoring of the potentials.

A.4 MUSCLE DISORDERS

A.4.1 Polymyositis A muscle disease in which there is muscular weakness. It is classified in four groups. In the first, there are muscular changes without involvement of skin. In the second group, skin changes are a feature, although muscular weakness is again dominant. In the third group, muscle changes occur as a feature of a predominantly connective tissue disorder. In the fourth group, polymyositis occurs in association with malignant disease.

A.4.2 Myasthenia Gravis This is a neuromuscular transmission disorder involving the myoneural junctions. Nerve impulses fail to induce normal muscle contraction. The disease is characterized by great muscular weakness (without atrophy) and progressive fatigability.

A.4.3 Neuropathy Any disease of the nerves.

A.4.4 Myopathy Any disease or abnormal condition of the striated muscle.

APPENDIX B

EMG RECORDING - INFORMATION SHEET

SUBJECT ..... Date .....

TITLE OF ANALOG TAPE .....

SIDE OF TAPE .....

REVOLUTION COUNTER START .....

1. SECTION END ..... COMMENTS

2. SECTION END ..... COMMENTS

3. SECTION END ..... COMMENTS

4. SECTION END ..... COMMENTS

5. SECTION END .....

COMMENTS

DIAGNOSIS

DOCTOR IN CHARGE .....

OPERATOR .....

DISA ELECTROMYOGRAPH

FREQ. LIMITS .....

SENSITIVITY .....

TAPE RECORDER

SPEED .....

CHANNEL NO. .....

COMMENTS

APPENDIX C

DOCUMENTATION ON THE INTERACTIVE GRAPHICS ROUTINE

INTERACT - A program to interactively select motor unit potentials from an EMG time series waveform.

C.1 Introduction

INTERACT is an interactive graphics program developed for use at the Adage Graphics Terminal, U.B.C. Computing Center. Its purpose is to allow the selection of a single motor unit potential from the EMG time series waveform which is stored in a file record. The potential may be enlarged on the screen and by means of a cursor the operator may select the beginning and end of a firing interval which can be extracted and stored in another file for future analysis. The program further allows the operator to 'store' the potential on the screen and thus other potentials may be matched, aligned to it and extracted.

C.2 Running the Program

The program is initiated by an MTS command:

```
$RUN    INTERACT    1=WAVEFORM    2=POTENTIALS
```

WAVEFORM and POTENTIALS are the names of two sequential files, the former containing the EMG waveform to be examined and the other will contain the extracted action potentials. The file WAVEFORM should have data in the form of 2 bytes/integer and 4096 bytes/record. The file POTENTIALS will contain one motor unit potential per record, each sample being a 2 byte integer.

The operator will be asked to

ENTER SCALING FACTORS IN F10.1 FORMAT:MIN THEN MAX.

These scaling factors are the minimum and maximum values to which all

records of displayed EMG data will be scaled to on the screen.

The operator will then be asked to:

ENTER RECORD NUMBER

This allows the program to keep count of the number of records that have been displayed on the screen. The number should be entered in I4 format; for example, usually the first record in a file will be displayed and the operator will enter 0001.

C.3 Controls

Fig. C.1 shows the layout of the function button and their assignments.

Fig. C.2 shows the variable control dials.

BUTTON 11

This will terminate the program, returning the operator to MTS. Issuing the MTS command $RESTART will restart the program at the same point as it was terminated. The input or output files may be redefined if desired at this stage by the command $RESTART 1=NEW WAVEFORM and/or 2=NEW POTENTIALS.

BUTTON 12

This button causes the Adage Computer to read the next sequential record in the file WAVEFORM into an input buffer. The entire record in the buffer is then displayed on the screen. When an end of file is encountered, the program will terminate.

It should be noted that operations of the other function buttons are performed on the input buffer.

BUTTON 13 - GROW

This will cause the points presently on the display screen to be rescaled and spread out; only about half of the original points (those points about the center), will be 'grown'. This causes no permanent

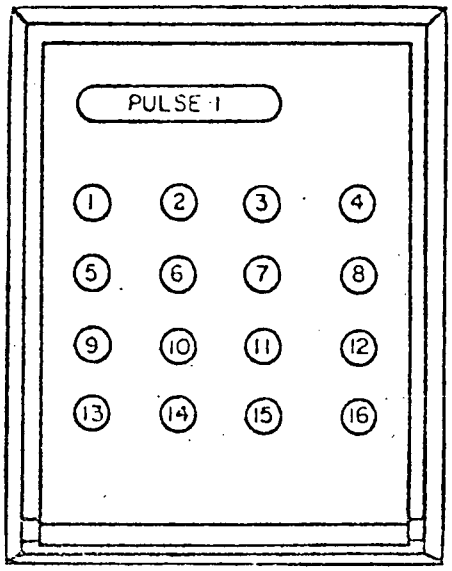


FIG. C.1 FUNCTION BUTTONS -LAYOUT AND ASSIGNMENT

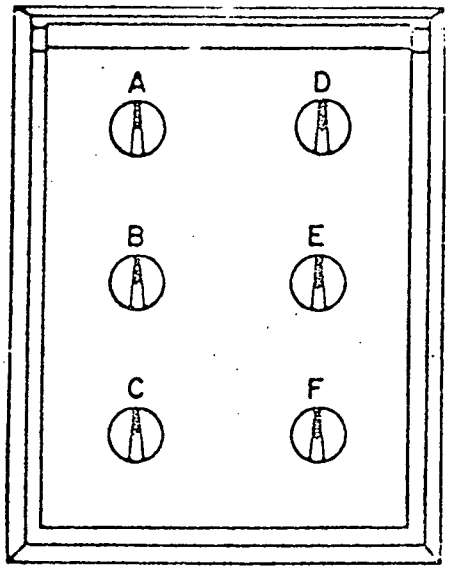


FIG. C.2 VARIABLE POTENTIOMETER CONTROL DIALS

changes in the values of the sample points in the record.

BUTTON 14 - SHRINK

This will double the number of points on the screen or, if all the points in a record are displayed, it reduces the step size between the points. No permanent changes in the values of the parts are produced.

BUTTON 15 - CENTER

This together with the x cross-hair (see later) allows points to be scrolled onto the screen from the left or right. It is normally used to center a selected motor unit potential on the screen (usually the peak of the potential is chosen as the center) before enlargement.

BUTTON 3

This enables the sample point nearest the x cross-hair to be chosen as the start of a range of points to be extracted and stored in the file POTENTIALS.

BUTTON 7

This selects the point nearest the x cross-hair to be the end of a range of points. The program will display on the terminal the number of sample points in the chosen range by

THE NUMBER OF SAMPLE POINTS = XXXX

The operator will be prompted to respond by

ENTER YOUR RESPONSE 1=YES, 2=NO.

Entering 1 will cause the chosen range of points to be extracted and stored in the next sequential record of file 2.

Entering 2 will allow the operator to reselect BUTTON 7 or alternatively reselect BUTTON 3 and BUTTON 7.

It should be noted that if BUTTON 7 is depressed before BUTTON 3 the program will respond

MUST USE BUTTON 3 FIRST.

BUTTON 1

This changes a point on the screen identified by the x cross-hair to a new coordinate identified by the y cross-hair. This feature may be used to mark the beginning and end of a range of points. To do this it must be used after BUTTON 3 and after BUTTON 7. The other function buttons are not required by the program.

C.4 Variable Control Dials

In Fig. C.2 control potentiometer A controls the position of the y cross-hair on the screen while A controls the position of the x cross-hair. The control potentiometer labelled B is used to control the brightness of the trace on the screen and E controls the aspect.

C.5 Foot Switches

There are 2 foot switch controls, one is used to 'store' the trace currently displayed on the screen. This is indicated by a proportionate increase of the trace brightness with each depression of the switch. This trace will remain on the screen while further records are displayed and any number of traces may be stored on the screen. The other foot switch will 'unfreeze' the trace and if it does not belong to the currently displayed record, it will disappear off the screen.

C.6 Additional Features

Once a motor unit potential has been identified, centered and 'stored' on the screen, the next selected potential may be matched to it. During this matching process, the operator may move the x cross-hair to the marked beginning of the trace and depress BUTTON 3. On depression of BUTTON 7 the program automatically adds the previously defined number of sample points to the beginning. The operator is still prompted for his

response; a 'No' means the number of sample points to be extracted to be redefined.

The program also keeps count of the number of records that have been displayed. The record number will appear on the terminal after the use of BUTTON 12.

The number of motor unit potential that are stored in file 2 is also displayed after a new potential has been extracted.

When a record is displayed on the screen, 12 additional points are added to the display. The first two points represent the maximum and minimum scale factors while the last ten points are set to zero, thus defining a zero baseline.

APPENDIX D

COMPUTER LISTING OF THE MOTOR UNIT POTENTIAL SIMULATION MODEL

```
C-----------------------------------------------------------------------
C
C ** THIS PROGRAM REPRESENTS A MODEL SIMULATING THE HUMAN MUSCLE ACTION
C    POTENTIAL**
C
C
C* DEFINITION OF INPUT PARAMETERS:
C  ZEE___NUMBER OF SINGLE MOTOR UNIT POTENTIALS TO BE FORMED
C  NFIBS___NUMBER OF SINGLE FIBERS IN AN ARRAY
C  NU___NUMBER OF POINTS ON THE X COORDINATE
C  DEV___STANDARD DEVIATION OF A GAUSSIAN DISPERSION OF FIBERS
C  JITTER___VALUE OF JITTER IN MICROSECONDS(SET TO A GAUSSIAN DISTRIBTION
C  NOTE... JITTER IS THREE TIMES THE STANDARD DEVIATION
C  AJITT___VALUE OF JITTER APPLIED TO AMPLITUDE
C  FIRST___INTIAL POINT ON X COORDINATE FOR THE POTENTIAL
C  NCYN___NUMBER OF CYLINDERS IN THE ARRAY
C  D(I)___DISTANCE FROM ELECTRODE OF CYLINDER I
C  N(I)___NUMBER OF FIBERS IN CYLINDER I
C
C* PARAMETERS IN PROGRAM:
C  SCALE =0.004 AS .1 MM IS EQUAL TO 25 MICROSECONDS
C  PATH(I)___AXIAL DISPERSION FOR ALL FIBERS
C  START(I)___JITTER VALUES FOR SINGLE FIBERS
C  THI(I)___SINGLE FIBER POTENTIAL VECTOR
C  VECT(I)___SUMMED SINGLE FIBERS POTENTIALS IN A MOTOR UNIT
C
C* OUTPUT
C
C  UNIT 1 WILL CONTAIN 'ZEE' MOTOR UNIT POTENTIALS
C*UPPER LIMITS _____100 MUP'S , 800 SINGLE FIBERS , 200 POINTS
C
C                          LAST UPDATE  7 APRIL 1976
C-----------------------------------------------------------------------
      REAL DIS(800),VECT(200),START(800),PATH(800)
      REAL THI(200),D(200)
      REAL JITTER
      INTEGER ZEE,N(200)
      INTEGER*2 LEN
      SCALE=0.004
C
C READ IN THE PARAMETERS
      READ(5,6) ZEE,NFIBS,NU,DEV,JITTER,AJITT,FIRST
6     FORMAT(I3/I3/I3/F10.3/F10.3/F10.3/F10.3)
      WRITE(6,401) ZEE
401   FORMAT(2X,'THE NUMBER OF MOTOR UNIT POTENTIALS GENERATED=',I3)
      WRITE(6,402) NFIBS
402   FORMAT(20X,'NUMBER OF FIBERS IN THE ARRAY=',I3)
      WRITE(6,403) NU
403   FORMAT(20X,'NUMBER OF POINTS CONSIDERED=',I3)
```

```
      WRITE(6,404) DEV
404   FORMAT(20X,'THE AXIAL DISPERSION=',F10.3)
      WRITE(6,405) JITTER
405   FORMAT(20X,'THE JITTER IN MICROSECONDS=',F10.3)
      WRITE(6,406) AJITT
406   FORMAT(20X,'THE AMPLITUDE JITTER IS SET AT ',F10.3)
      WRITE(6,407) FIRST
407   FORMAT(20X,'THE FIRST POINT CONSIDERED IS ',F10.3)
      IF(DEV.NE.0.0) GO TO 2
C
C  SET PATH TO ZERO IF NO AXIAL DISPERSION OF FIBERS
      DO 54 I=1,NFIBS
54    PATH(I)=0.0
      GO TO 12
C
C APPLY GAUSSIAN AXIAL DISPERSION (IF DEV IS 0.0 __NO DISPERSION)
2     P=123.4567
      Z=RANDN(P)
      DO 94 I=1,NFIBS
      F=FRANDN(D)
94    PATH(I)=F*DEV
12    CONTINUE
C
C DEFINE THE FIBER ARRAY
C
      READ(5,8) NCYN
8     FORMAT(I3)
      DO 901 I=1,NCYN
      READ(5,191) N(I)
191   FORMAT(I3)
      READ(5,902) D(I)
C
C TEST IF FIBER IS TOO CLOSE TO POINT SOURCE ELECTRODE
901   IF(D(I).LE. 0.08) D(I)=D(I)+0.08
902   FORMAT(F10.3)
      INDEX=0
      DO 300 ICYN=1,NCYN
      NEND=N(NCYN)
      IF(ICYN.GT.1) INDEX=INDEX+N(ICYN-1)
      DO 200 I=1,NEND
      DIS(I+INDEX)=D(ICYN)
200   CONTINUE
300   CONTINUE
121   CONTINUE
C
C INTIALIZE THE NUMBER OF MOTOR UNIT COMPUTATIONS
      NCOUNT=0
      LEN=4*NU
39    CONTINUE
C
C APPLY GAUSSIAN JITTER
      IF(JITTER.NE.0.0) GO TO 3
C
C  IF NO JITTER THEN EACH POTENTIAL HAS SAME STARTING POINT
      DO 55 I=1,NFIBS
```

```
55      START(I)=FIRST
        GO TO 44
3       T=SCLOCK(0.0)
        Z=RANDN(T)
        DO 44 I=1,NFIBS
        F=FRANDN(D)
        STAND=JITTER/3.
C
C  ASSUME JITTER LIES WITHIN +/-3 STANDARD DEVIATIONS
        START(I)=F*STAND
C
C SCALE TO MILLIMETERS
1       START(I)=START(I)*SCALE
C
C FIND INTIAL POINT OF COMPUTATION WITH RESPECT TO 'FIRST'
13      CONTINUE
        START(I)=START(I)+FIRST
44      CONTINUE
C
C INTIALIZE VECT(I) TO ZERO
C VECT CONTAINS SUMMED CONTRIBUTIONS FROM SINGLE FIBERS
        DO 30 I=1,NU
30      VECT(I)=0.0
C
C APPLY JITTER AND DISPERSION
C NOTE......ONLY THE JITTER WILL VARY IN EACH DIFFERENT MUP
112     DO 21 I=1,NFIBS
        START(I)=START(I)+PATH(I)
        CALL FIBER(DIS(I),START(I),THI,NU,AJITT)
C
C SUM UP SINGLE FIBER CONTRIBUTIONS
        DO 29 J=1,NU
29      VECT(J)=VECT(J)+THI(J)
21      CONTINUE
        NCOUNT=NCOUNT+1
        IF(NCOUNT.GT.ZEE) GO TO 4
C
C CHECK IF THE DESIRED NUMBER OF MUP'S HAVE BEEN COMPUTED
C STORE THE POTENTIAL IN UNIT 1
        CALL WRITE(VECT,LEN,0,LNR,1)
        GO TO 39
4       CONTINUE
        STOP
        END
C
        SUBROUTINE FIBER(DIS,START,THI,NUMB,AJITT)
C THIS SUBROUTINE COMPUTES THE SINGLE FIBER POTENTIAL
C THIS SINGLE FIBER POTENTIAL IS BASED ON THE DIPOLE MODEL
C OF ROSENFALCK (1969)
        REAL THI(NUMB)
        R=DIS
        S=0.25
C
C DISTANCE BETWEEN THE POLES =2*S
        IF(AJITT.NE.0.0) GO TO4
        AMP=1.0
```

```
      Z=RANDN(T)
      AMP=F*AJITT
12    CONTINUE
      Z=START
      DO 15 I=1,NUMB
      A=AMP/(SQRT((Z+S)**2+R**2))
      B=AMP/(SQRT((Z-S)**2+R**2))
      THI(I)=A-B
C
C INVERT THE POTENTIAL (DOWNWARDS +VE)
C  THIS IS THE CONVENTION IN ELECTROMYOGRAPHY
      THI(I)=-THI(I)
15    Z=Z+0.1
C
      GO TO 12
C
C     APPLY AMPLITUDE JITTER
4     T=SCLOCK(0.0)
C INCREMENT BY .1 MM
      RETURN
      END
```

APPENDIX E

COMPUTATION OF THE AVERAGED POTENTIAL, THE VARIANCE AND THE PLOTTING OF RESULTS

```
C-----------------------------------------------------------------------
C**  THIS PROGRAM COMPUTES THE AVERAGE SIMULATED POTENTIAL,
C    THE VARIANCE,AND PLOTS THE RESULTS  **
C
C
C*  INPUT PARAMETERS
C  NAVG__NUMBER OF POTENTIALS
C  UNIT 1___CONTAINS THE MOTOR UNIT POTENTIALS
C*  OUTPUT
C  UNIT 6___VARIANCE VALUES ARE PRINTED  ON THIS UNIT
C
C  UNIT 7___WILL CONTAIN THE AVERAGE POTENTIAL
C
C* UPPER LIMITS ---200 MOTOR ACTION POTENTIALS
C
C                              LAST UPDATE =5 APRIL 1976.
C
C-----------------------------------------------------------------------
C  THIS PART OF THE PROGRAM COMPUTES THE AVERAGE MOTOR UNIT POTENTIAL
C
      REAL Y(200),X(200),AVG(200),VECT(200)
      INTEGER*2 LEN
      READ(5,290) NAVG
290   FORMAT(I3)
C
C INTIALIZATION
      NCOUNT=1
      DO 66 I=1,200
66    Y(I)=0.0
C
C  READ INTO ARRAY X THE POTENTIALS TO AVERAGED
11    CALL READ(X,LEN,0,LNR,1,&501)
      L=LEN/4
      DO 4 I=1,L
C
C SUM THE POTENTIALS
4     Y(I)=X(I)+Y(I)
      NCOUNT=NCOUNT+1
      IF(NCOUNT.LE.NAVG) GO TO 11
      F=NAVG
C
C FIND AVERAGE
      DO 23 I=1,L
23    Y(I)=Y(I)/F
C
C SAVE THE AVERAGE
      DO 24 I=1,L
24    AVG(I)=Y(I)
```

```
C
C   WRITE AVERAGED POTENTIAL INTO UNIT 7
      CALL WRITE(AVG,LEN,0,LNR,7)
C
C PLOT THE AVERAGE
      CALL DISP(Y,L)
C
C   THIS PART OF THE PROGRAM COMPUTES THE VARIANCE
C
C INTIALIZATION
      NCOUNT=0
      DO 14 I=1,200
14    VECT(I)=0.0
C
C COMPUTE THE VARIANCE
      REWIND 1
C
C   READ INTO ARRAY X THE POTENTIALS
3     CALL READ(X,LEN,0,LNR,1,&501)
      DO 5 I=1,L
5     VECT(I)=VECT(I)+(AVG(I)-X(I))**2
      NCOUNT=NCOUNT+1
      IF(NCOUNT.LT.NAVG) GO TO 3
      DO 29 I=1,L
29    VECT(I)=VECT(I)/F
C
C   WRITE VARIANCE ON UNIT 6
      WRITE(6,99) (VECT(I),I=1,L)
99    FORMAT(2X, 5G10.4)
C
C PLOT THE VARIANCE
      CALL DISP(VECT,L)
      CALL PLOTND
      STOP
501   STOP 501
      END
      SUBROUTINE DISP(Y,L)
C
C THIS SUBROUTINE CONTAINS THE PLOTTING ROUTINES
      REAL TIME(200),Y(200)
      TIME(1)=0.0
      DO 25 I=2,L
25    TIME(I)=TIME(I-1)+0.1
      CALL SCALE(Y,L,5.,XMIN,DX,1)
      CALL SCALE(TIME,L,6.,YMIN,DY,1)
      CALL AXIS(0.,0.,'AMPLITUDE',9,5.,90.,XMIN,DX)
      CALL AXIS(0.,0.,'DIST(MILLIMETERS)',-17,6.,0.,YMIN,DY)
      CALL PLOT(TIME(1),Y(1),+3)
      CALL LINE(TIME,Y,L,1)
      CALL PLOT(12.0,0.0,-3)
      RETURN
      END
```

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